THE INFLUENCE OF WHEELCHAIR MECHANICAL PARAMETERS AND HUMAN PHYSICAL FITNESS ON PROPULSION EFFORT

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THE INFLUENCE OF WHEELCHAIR MECHANICAL PARAMETERS AND HUMAN PHYSICAL FITNESS ON PROPULSION EFFORT

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This thesis is dedicated to those wheelchair users who willingly participate in this research to advance our understanding of how wheelchair parameters and human physical fitness influence wheelchair propulsion.

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LIST OF SYMBOLS AND ABBREVIATIONS

SCI	Spinal Cord Injury
MWU	Manual Wheelchair Users
TBI	Traumatic Brain Injury
DW	Drive Wheel
ADL	Activity of Daily Living
KE	Kinetic Energy
RPE	Rating of Perceived Exertion
Net effort_Tile; Net effort_Carpet	Net propulsion effort on tile or carpet
XPOS	The shoulder position
F_{iso}	Propulsion strength
Decel_St	Decelerations in straight direction
Decel_Turn	Decelerations in turning direction
Mass_sys	System Mass
%DW	The percentage of system mass on drive wheels

SUMMARY

For wheelchair users, the ease of maneuvering a wheelchair is crucial for their mobility and participation in their communities, thus improving their quality of life. From a mechanical design standpoint, the major factors influencing propulsion efforts are inertia and frictional energy loss. On a wheelchair with greater inertia and/or greater frictional loss, a user needs to exert greater instantaneous forces and metabolic costs while completing a maneuver. Greater propulsion effort can lead to difficulty in achieving the desired speed and a higher probability of fatigue over long bouts of mobility. The majority of wheelchair studies that attempt to evaluate propulsion efforts across wheelchair configurations examines long and steady propulsion. However, the results of these studies cannot represent performance during daily maneuvers, which include changes in speed and direction. Physical fitness was proved to be related to health status and exercise performance. In addition, the biomechanical characteristics of the user were widely studied in the impact of wheelchair maneuvers. However, it is still unknown how these operator factors would influence wheelchair propulsions together. Therefore, the overall objective of the study is to define the relative influence of mechanical wheelchair parameters as well as individual physical and biomechanical variables on propulsion efforts during over-ground maneuvers.

To meet the overall objective, we define three specific aims. The first is to develop and validate a test that quantifies the impact of wheelchair configurations on frictional energy loss, particularly loss related to turning trajectories. The second is to develop and validate a testing protocol designed to measure maximum propulsion

strength, which will test subjects in a realistic condition – while seated in their wheelchairs. The third is to identify the impact of the mechanical parameters of wheelchairs as well as the physical and biomechanical variables of operators on propulsion efforts during over-ground maneuvers. Mechanical parameters include both inertial and frictional measurements. Operator factors include shoulder position, propulsion strength, and aerobic capacity. Biomechanics studies commonly control for these operator factors, which this study will analyze to evaluate the physical fitness of wheelchair users. To evaluate the performance of daily maneuvering, we designed a repeatable maneuver consisting of several momentum changes. Because of the breadth of wheelchair configurations and variance in user physical capacity, it is necessary to define the effects of wheelchair configurations and user fitness on propulsion with a systematic approach.

We found that the shoulder position and weight distribution had a significant influence on the frictional energy loss and propulsion efforts. However, aerobic capacity and muscle strength had less influence on daily wheelchair maneuver. Clinicians can use our finding, which covers wheelchair designs and human fitness, to select equipment and prescribe exercise to wheelchair users. Manufacturers can also improve their wheelchair design by understanding the importance of shoulder position and weight distribution.

CHAPTER 1

INTRODUCTION

1.1 Background and significance

In the United States, an estimated 1.5 million people use manual wheelchairs [1]. Over 60% of wheelchair users have reported a poor or fair health status, about twice as high as those who do not use mobility devices [1]. Mobility is crucial for full participation in educational, vocational, and personal activities, but wheelchair users move about much less than people who ambulate, constituting a significant "mobility disability" [2-5], which can adversely impact one's quality of life. Conversely, increased mobility is associated with better outcomes in health, activity, and participation [6, 7]. Since maneuvering a wheelchair requires the upper limbs to produce repeated and forceful movements, the amount of effort required for maneuvers has an obvious effect on the amount of mobility in everyday life.

Compared to ambulating, propelling a wheelchair is less energy efficient; therefore, the amount of work performed by the user does not efficiently translate into distance traveled. In contrast to the efficiency of walking of about 15% [8, 9], the gross mechanical efficiency of push-rim propulsion seldom exceeds 10% [10, 11]. Propelling a wheelchair with the greater frictional loss, the user must exert greater effort while completing a maneuver. Greater effort can lead to a higher probability of fatigue over long bouts of mobility. Over time, the accumulation of inefficient activity can potentially cause upper extremities injuries. According to Boninger's et al. shoulder MRI [12] and nerve (median and ulnar nerve) function [13] studies, they found that greater propulsion forces directed toward the wheelchair axle could increase the risks of MRI abnormalities and long, smooth strokes may benefit nerve health.

These issues of mechanical efficiency and effort have motivated a substantial body of research targeted towards improving wheelchair propulsion at the component level. We can roughly group such work can be into 1) studies of mechanical systems and components, and 2) biomechanical studies of individuals propelling wheelchairs. Studies related to mechanical systems have focused on friction and inertial influences and produced useful information on rolling resistance as a function of the wheelchair [14, 15], the caster size [16], the tire design, and frame material [17-19]. Many other studies of propulsion behaviors have focused on neuromuscular responses [20, 21], biomechanics [22-24], and mechanical efficiency [25-27] with respect to various wheelchair configurations or designs.

<u>The mechanical and component features of the wheelchair system</u>

With respect to the wheelchair, users must apply torque to overcome resistive loss resulting from friction, which exists in all wheeled vehicles. Resistive energy loss is largely the result of rolling resistance, bearing resistance, drive wheel and caster scrub, and other frictional factors such as drag and frame flexibility [28, 29]. Because energy loss is directly related to users' muscular and metabolic effort [30], researchers have characterized such loss using a variety of techniques.

Resistive loss of wheelchair tires and entire wheelchair systems have been measured experimentally using a treadmill [17, 26] and a dynamometer [31]. These approaches can isolate resistive loss resulting from a single component (e.g., drive wheels), but they are limited to testing only the rolling surfaces of the respective equipment. Freewheeling roll-down tests [32, 33] have been used to measure resistive loss of the entire wheelchair system and study the effect of various rolling surfaces. These studies have proposed useful approaches for describing resistive loss during a straight trajectory, but their results did not apply to evaluating resistive loss resulting from tire scrub or caster shimmy in turning [16, 29, 34]. Therefore, **my first aim was to quantify the impact of various wheelchair configurations on overall frictional loss, especially during turning**. Without using component-level analysis, which does not reflect the complex interactions among wheelchair components on a systems level, **my approach described the overall resistive loss that translates into clinically useful knowledge**.

<u>The physiological response and biomechanics of wheelchair propulsion</u>

The quality of physical performance is influenced by an individual's capacity for maximal energy output (i.e., maximal aerobic and anaerobic outputs), muscular strength, coordination/economy of movement, psychological factors, and environment **(Figure 1)**. Many types of physical activities require a combination of several of these factors to achieve exceptional performance. In the case of wheelchair propulsion, cardiorespiratory function, muscular performance, propulsion economy, and environment are critical factors and are the focus of investigation in this study. In contrast with healthy populations, wheelchair users' aerobic capacity and muscle strength are both significantly impaired by their injury or disease. For example, a spinal cord injury (SCI) directly alter the voluntary control of major muscle mass. For lesions above the sixth thoracic vertebra

(T5), the heart rate is also affected, thus lowering cardiovascular response during maximal exercise. Besides physical impairment, propulsion economy depends on not only muscle coordination and movement skills, but also on wheelchair seating. It is widely agreed upon that wheelchair users can improve their propulsion economy by choosing a well-fit wheelchair because of biomechanical benefits. In addition, the environment is characterized by the machine (wheelchair) and task (including surface/grade and trajectory) that would influence wheelchair propulsion.

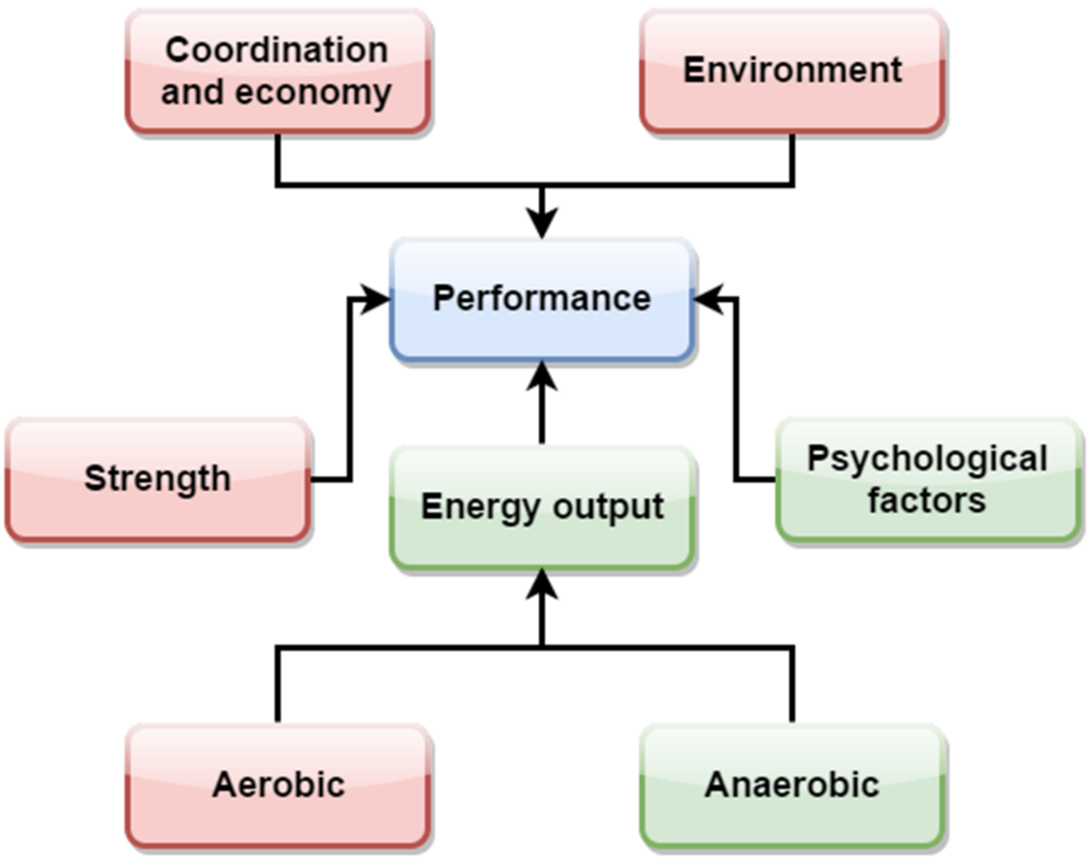


Figure 1 Factors that contribute to physical performance [35]

The mechanical characteristics of wheelchair configurations are related to the maneuverability of the wheelchair. Biomechanic research has focused on external influences that affect propulsion effort. External factors, such as axle position [23, 36-38], surface type [22, 39], tire type [31], and inflation [19], have been shown to influence propulsion effort. Operator factors, such as muscle strength and cardiorespiratory fitness, are also believed to influence exercise performance for people with SCIs [40-43]. For measuring muscle strength, most SCI studies either used modified ergometers [43, 44] or positioned subjects in sitting/lying postures [45-49]. However, these approaches are limited to measuring the strength of certain muscle groups instead of directly evaluating the strength of users while they are propelling a wheelchair. Although Janssen [43] designed a stationary wheelchair ergometer to measure torques and forces applied to push-rims, subjects were still restricted to using an experimental chair to do a push task. Therefore, my **second aim is to evaluate propulsion strength in realistic conditions by using users' wheelchairs.**

Depending on the extent of disability, aerobic [50] capacity can vary across injury types and levels. The performance of wheelchair propulsion also depends on their functional capacity [51, 52]. Dicarlo [53] found that improvements in cardiopulmonary function were positively correlated with propulsion endurance and physical work capacity. These studies have shown that the aerobic capacity of MWU not only reflects subjects' fitness but also influences propulsion performance. In addition, the relative distance from the shoulder to the axle position has been shown to influence the biomechanics of propulsion [22, 36]. Boninger et al. [24] found that the horizontal position of the axle with respect to the shoulder positively correlated with push force and frequency but negatively correlated with the push angle at fixed speeds. By definition, push angle is the number of degrees that the hand applies a propulsion force on the pushrim. During start-up propulsion, compared to subjects with a long distance between their shoulders and the axle position on a horizontal plane, subjects with a short distance exhibited stronger performance at acceleration, higher stroke frequency, but decreased shoulder range of motion [36]. Although axle position was clinically suggested to move as forward as possible to provide a better maneuverability, forward axle position could also decrease stability. According to a previous literature review in biomechanics, these wheelchair studies highlighted the importance of considering the impact of physiological and biomechanical factors on wheelchair propulsion.

Although the potential influence of mechanical and biomechanical characteristics on propulsion performance has been suggested, evidence has been limited to measuring expected steady-state propulsions instead of over-ground maneuvers. According to Sonenblum's et al. manual wheelchair study [3], the median bout of weekly activity lasted 21 seconds and traveled 8.6 m at 0.43 m/s. In addition, 85% of recorded bouts lasted less than one minute and traveled less than 30 meters. Research design in steady-state propulsion may result in two major limitations across wheelchair configurations. One limitation is that long durations of propulsion do not reflect actual wheelchair use in daily life [3]. Another limitation is that methodologies using steady-state propulsion do not include the impact of inertial changes on propulsion efforts, or they do not consider the frictional energy losses while turning [54]. As a result, studying propulsion with regard to only straight trajectories has limitations to fully characterizing the inertial and frictional influence of the propulsion effort.

The objective of this project was to evaluate **the impact of wheelchair configurations, shoulder position, and fitness status (including propulsion strength**

and aerobic capacity) on propulsion efforts during over-ground maneuvers. We believed that building a model is helpful to identify the related influence of mechanical, biomechanical, and physical factors on wheelchair propulsion (**Figure 2**). These results would provide a useful reference those choosing a proper wheelchair and training wheelchair users.

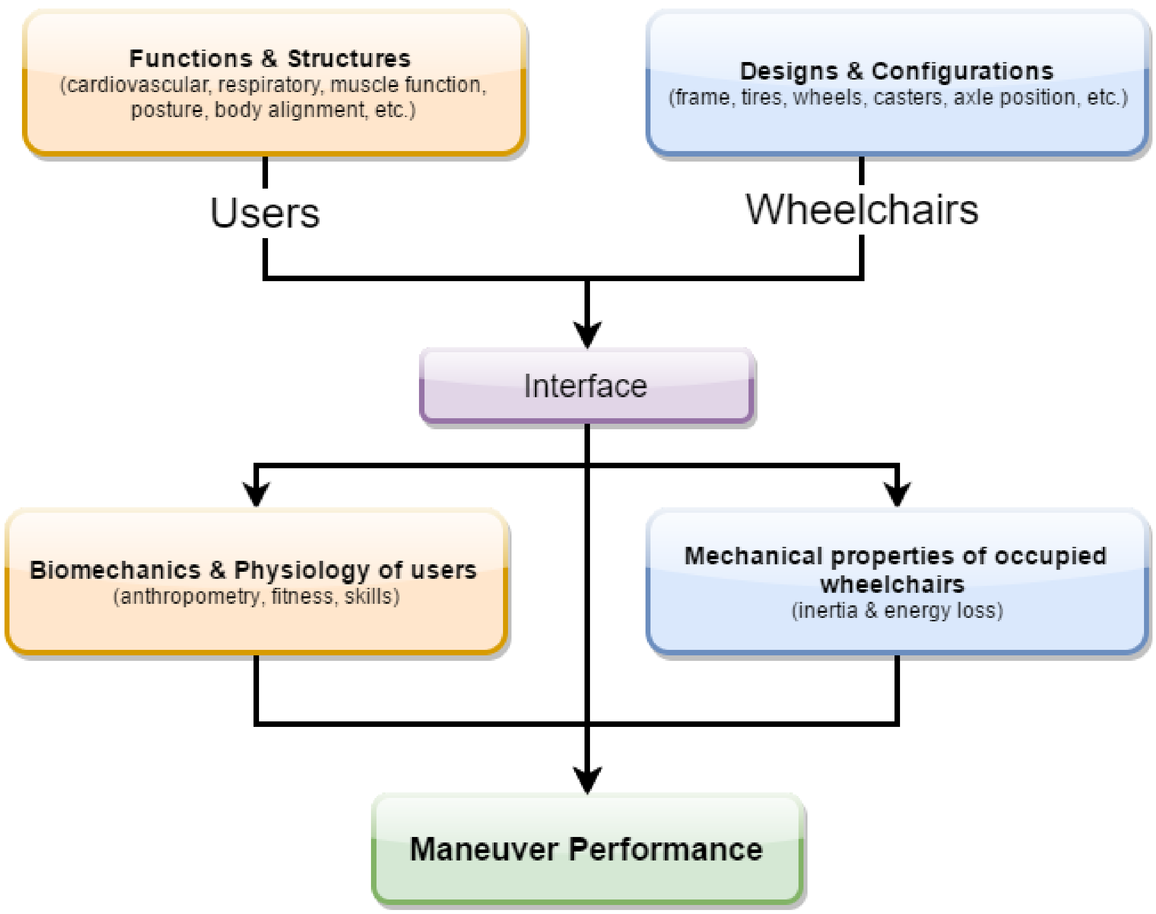


Figure 2 Factors that influence wheelchair maneuvers

1.2 Specific Aim #1: To quantify the impact of wheelchair configurations on frictional energy loss in turning trajectories

Turning is one of the most common propulsion tasks in daily activity [3]. Evaluating energy loss, particularly during the turning maneuver, is essential for characterizing the frictional parameters of wheelchairs. However, no studies have developed a protocol to study the friction of tire scrub in a systematic approach, Aim#1.1 was to develop a reliable and valid field test to measure <u>overall energy loss</u> during straight and turning trajectories. Aim#1.2 also evaluated the influence of mass and weight distribution, two major inertia properties of wheelchair designs that impact friction. We <u>hypothesize</u> that changing weight distribution impacts frictional energy loss in straight and turning differently. <u>The results of Aim#1 will provide insight into the influence of weight distribution and surface as a determinant of frictional energy loss.</u>

- Aim #1. 1: Validate the coast-down approach to evaluating frictional energy loss during straight and turning trajectories
- Aim #1. 2: Identify the frictional energy loss across wheelchair configurations

1.3 Specific Aim #2: To develop and validate a testing protocol for measuring maximum propulsion strength

Maximum isometric contraction is commonly used to evaluate muscle strength [43, 45]. An accepted procedure for quantifying the maximum activity of each muscle group, is strength testing in predefined postures [43-49]. However, while these measures are valid for upper extremity strength in isolated planes of motion, the functional propulsion strength may differ when a person is seated in a wheelchair. In clinical situations, both muscle strength and external factors such as seat position, hand-rim design, and propulsion skills influence propulsion [52]. Therefore, Aim # 2.1 was to develop a unique approach in which strength is measured under realistic conditions—while a subject is seated in a wheelchair. Further, Aim #2.2 will evaluate test-retest reliability and validity by comparing to gold standard afterward. The hypothesis is that propulsion strength, strength measurements of users seated in a wheelchair, will strongly correlate with but differ from those using a Kin-Com dynamometer system. The results of Aim 2 will provide an valid approach to measuring upper-extremity strength specifically related to the task of propulsion.

- Aim #2. 1: Develop a testing procedure for measuring maximum isometric propulsion strength while subjects are seated in their wheelchairs
- Aim #2. 2: Evaluate the repeatability and validity of the test protocol to measure maximum propulsion strength

1.4 Specific Aim #3: To develop a regression model that identifies the impact of users' physical factors and wheelchairs' mechanical properties on propulsion efforts during over-ground maneuvers

Measuring metabolic effort during over-ground maneuvers is complicated because changes in speed and direction will affect propulsion effort throughout the maneuver. However, over-ground maneuvers that consist of changes in momentum are much more representative of everyday mobility and more accurately reflect the propulsion effort required for maneuvering wheelchairs [3]. Therefore, Aim #3.1 was to develop a repeatable over-ground maneuver, which includes the features of acceleration, deceleration, and turns.

Because of variable levels of impairment in the cardiovascular and/or muscular systems of wheelchair users, their diverse functionality results in variable levels of physical fitness [50-52, 55]. Physical fitness, which includes aerobic capacity and muscle strength, is commonly used to evaluate the fitness status of MWU and associated with propulsion performance [56]. In addition to the physical fitness, the axle position has been shown to impact propulsion biomechanics [23, 24, 57]. Therefore, this study will evaluate the related influence of physical fitness and biomechanical parameters on propulsion effort. Because of any change in wheelchair configuration influences inertia and/or the friction of the wheelchair system, this study will also evaluate the related influence of mechanical parameters on propulsion effort. We hypothesize that overall propulsion effort is a function of both the physical and biomechanical variables of the wheelchair user and the mechanical parameters of the wheelchair. Specifically, we hypothesized that mechanical parameters of wheelchairs, such as greater mass/inertia and friction, would require greater metabolic efforts while maneuvering wheelchairs. We also hypothesized that the physical fitness and biomechanical parameters of operators, such as greater muscle strength, better aerobic capacity, and better ergonomic position, can improve propulsion performance by reducing metabolic cost. According to the related influences of physical variables and mechanical parameters, clinicians can use the results of Aim 3 to prescribe an optimal wheelchair and manufacturers to improve wheelchair design.

Aim #3.1: To develop and evaluate a repeatable propulsion course that highlights changes in speed and direction

Aim #3.2: To characterize wheelchair users' physical and biomechanical variables and determine their relationship to propulsion effort

Aim #3.3: To identify the mechanical parameters of the wheelchair and determine their relationship to propulsion effort

Aim #3.4: To evaluate the combined impact of operator and mechanical factors on propulsion effort

CHAPTER 2
TO QUANTIFY THE IMPACT OF WHEELCHAIR CONFIGURATIONS ON FRICTIONAL ENERGY LOSS IN TURNING TRAJECTORIES

2.1 Aim # 1.1: Validate the coast-down approach to evaluating frictional energy loss related to straight and turning trajectories

When propelling a manual wheelchair (MWC), users apply torque to drive wheels (DWs) in order to achieve a specific maneuver. The magnitude of this propulsion torque is dependent on the kinematics of the maneuver and the characteristics of the user-wheelchair system. On a chair with greater frictional resistance, the user needs to exert greater instantaneous force and total effort while completing a maneuver [29, 30, 58]. Greater effort can lead to difficulty in achieving desired speeds, a higher probability of fatigue over long bouts of mobility, and difficulty negotiating inclines. Over time, the accumulation of this greater effort can increase the potential for injury in the upper extremities, a complication in manual wheelchair users that has been well-described for many years [12, 59, 60].

Improving wheelchair propulsion has motivated a substantial body of research, including studies on mechanical systems and components with regard to inertia and resistive loss [14, 15]. When moving over ground, resistive energy losses occur due to a combination of rolling resistance, bearing resistance, tire scrub, and other frictional factors such as drag and frame flexion [28, 29]. Researchers have used a variety of techniques to characterize these losses.

The resistive losses of the wheelchair tires and entire wheelchair systems have been measured experimentally using component-level analysis [61], treadmill [17, 26], dynamometer [31], and roll-down tests [15, 32, 33, 62]. Kauzlaurich and Thacker [61] demonstrated via both component and system analysis that rolling resistance is related to tire material, wheel radius, profile radius, and load on the wheel base. For a typical polyurethane solid tire and a 200-lb occupant, they calculated a rolling resistance of 13.4 N [61]. Gordon et al. [17] and Kwarciak et al. [31] measured rolling resistances of wheelchair DWs using a treadmill and dynamometer, respectively. These approaches can isolate resistive losses due to a single component (e.g., DWs) but were limited to testing on only the rolling surfaces of the respective equipment. Freewheeling roll-down tests have been used to measure resistive losses of the entire wheelchair system and to study the effect of different rolling surfaces. For example, Hoffman et al. [32] used three timing sensors to measure decelerations on different wheelchair frames, wheel combinations, and ground surfaces. Both Bascou et al. [33] and Sauret et al. [62] developed a mechanical model of MWC deceleration to describe the impact of mass distribution and the ground surface on rolling resistance properties. These studies concluded that wheelchair configurations can significantly affect the rolling resistance [32, 33, 62]. While these coast-down tests have been capable of characterizing overall rolling resistance, the methods have been constrained to straight trajectories. As a result, they are unable to capture the resistive forces from tire scrub associated with wheelchair turning maneuvers.

During turning maneuvers, the finite tire surface contacting the ground surface distorts as the wheels turn relative to the ground. This distortion is accompanied by a restoring force termed sideslip friction or scrub torque. The magnitude of this force is dependent on many properties, such as the load applied, contact area, and sideslip angle [63]. Sideslip angle is the angle between the wheel orientation and the direction of its velocity, and is typically small for both casters and DWs during steady-state fixed-radius turning. However, during instances where the turning radius changes and the casters swivel with respect to the wheelchair frame, sideslip angles of the casters become very large, elevating scrub torque substantially. Therefore, in all instances of turning, the scrub torque of the casters and DWs contribute to the overall resistive force.

Kauzlarich et al. [34] developed a wheel model to evaluate different caster designs on caster shimmy and turning resistance. The turning resistance of casters was quantified by measuring caster turning torque while applying normal loads via a drill-press. By constraining one of the DWs to pivot without rolling, system-level turning resistance was further evaluated by measuring the pivot torque necessary to initiate turning. The results have shown that casters with grooved tires (0.5") had 10% greater turning resistance than casters with un-grooved tires [34]. In addition, an estimated 7 N-m is required to initiate turning when 300N are loaded onto each DW [34]. Frank and Able [16] also studied turning resistance by measuring the torque required to pivot casters with diameters between 10 and 20 cm. Over a load range of 50 N to 300 N, turning resistance for a caster spanned a range from 0.3 N-m to 2.8 N-m [16]. Caspall, et al. [29] used these results to estimate that 8.75 N-m is needed to swivel wheelchair casters loaded to 100N, which was based upon a 20/80% caster-to-drive wheel weight distribution for a 90 kg user and a 12 kg wheelchair. These results document the resistive losses that occur when casters swivel and underscore the need to measure resistive losses of a wheelchair system during turning maneuvers in addition to straight trajectories. However, these papers have yet to quantify the impact of different wheelchair configurations and designs on overall resistive torque while turning.

In daily life, MWC users tend to move in relatively short bouts of movements that include turning [3]. Therefore, it is important to characterize resistive loss during both straight and turning maneuvers, and how these losses vary across different wheelchair designs and configurations. **Figure 3** illustrates frictional forces that must be overcome in both straight and turning trajectories. The frictional loss during straight trajectories is due to the rolling resistance of casters and wheels. A fixed-wheel turn involves both rolling resistance and tire scrub torque, so tire scrub torque is the distinguishing component between the trajectories. In Aim # 1.1, we measured rolling resistance related to straight trajectories based upon the deceleration profiles. In addition, we took turning maneuvers into account when characterizing frictional loss that includes tire scrubs.

Coast-down during straight trajectory **Coast-down during a fixed - wheel turn**

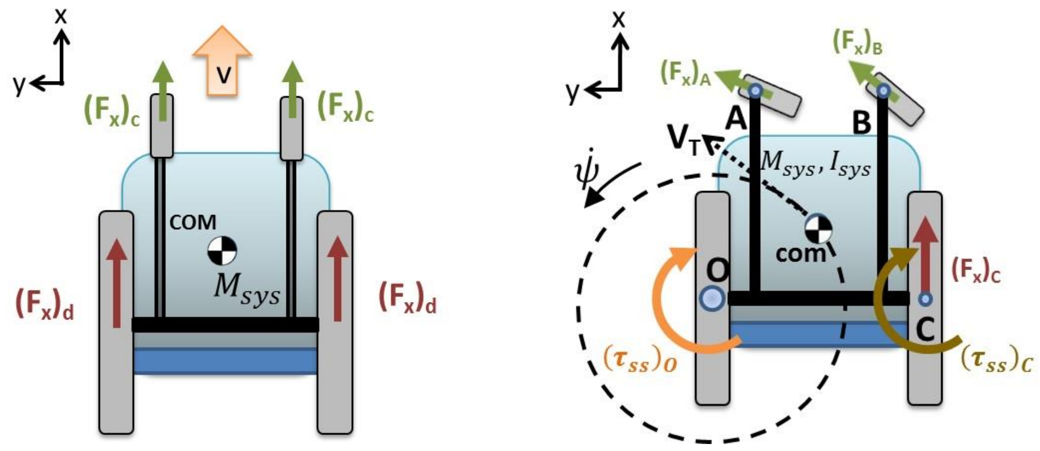


Figure 3 Model of straight motion and fixed-wheel turns with rolling resistance forces, F, and tire scrub torques, τ. CoM: center of mass; VT: tangential velocity.

2.1.1 Methods

<u>Test equipment and Instrumentation</u>

Test mannequin. A76 kg ISO 7176-11 dummy **(Figure 4**) was used as the wheelchair occupant during straight and fixed-wheel coast-down tests. A mannequin was chosen to avoid the confounding factors of body movement and postural changes that could be present when using human subjects [64]. This mannequin was configured to have a similar fore-aft location of the dummy center of mass as the new ISO standard (220mm from the dummy's back support reference plane) [65].

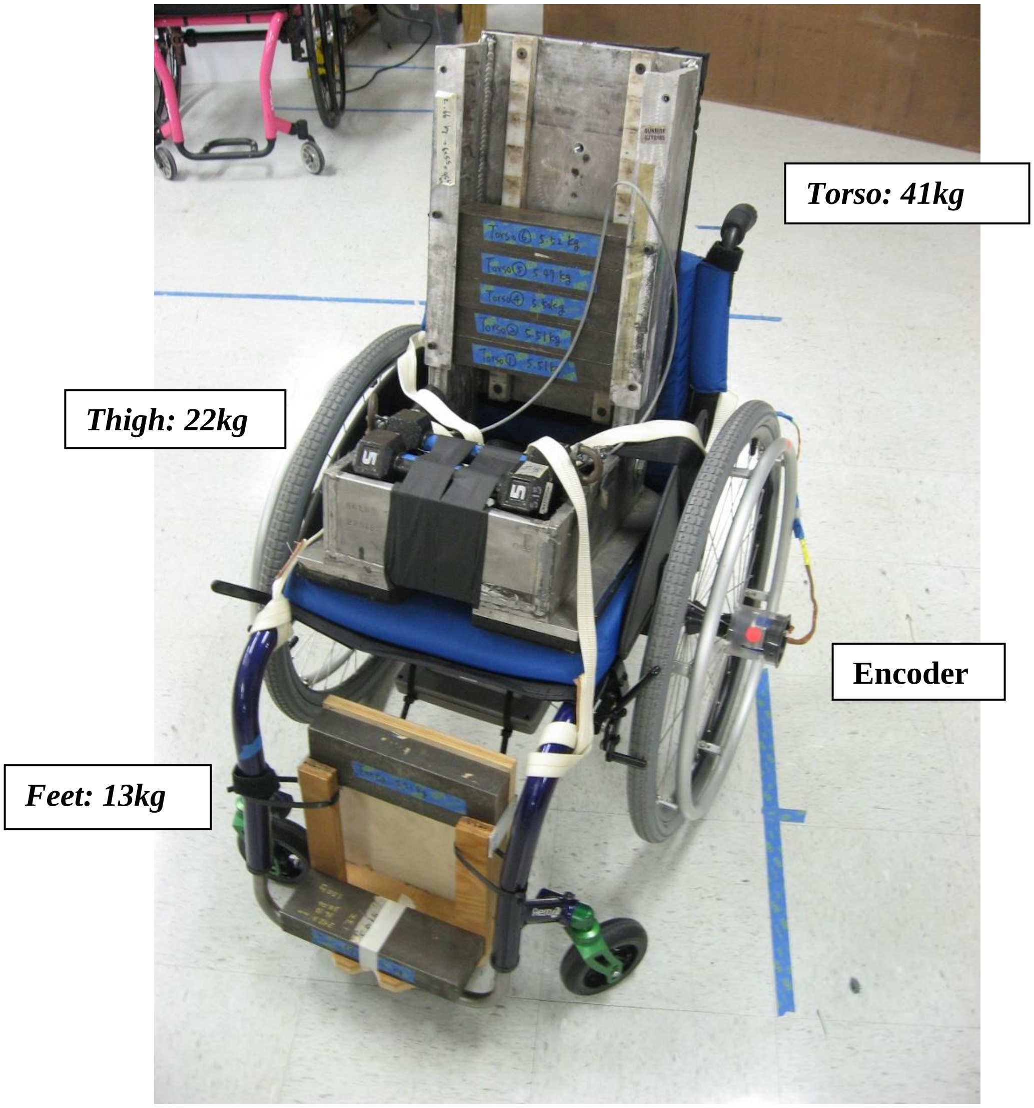


Figure 4 The setup of the testing wheelchair for coast-down study

Test surface. All coast-down tests were conducted on an indoor surface comprised of 12”×12” linoleum tiles. The standardized coefficient of kinetic friction of the tile surface was found to be 0.54 using RESNA’s test procedure to characterize surfaces during wheelchair testing [66].

System and wheel inertia. To calculate the resistive forces that decelerate the rolling wheelchair, inertial properties of the loaded wheelchair, wheels, and casters are required. Mass, yaw inertia of the wheelchair (I_G), and the location of the center of mass (CoM) were measured experimentally using a device called the iMachine [67]. The device consists of a turntable mounted to a single axle. Load cells mounted on the turntable measure the mass and CoM of the wheelchair. By measuring the CoM distance from the rear axle becomes possible to calculate the caster-to-drive wheel load distribution of the wheelchair. The detailed calculation of weight distribution was described in Aim 3 data analysis section. An encoder measures the rotation of the turntable, whose oscillations are damped by a spring of known stiffness, allowing calculation of rotational inertia from measured natural frequency. The rotational inertia of drive wheels and casters were measured using a system based on the established Trifilar Pendulum [68, 69], which measures the rotational inertia based on the mass and the frequency of oscillation.

Wheelchair encoders. Angular velocities (ω) of the DWs were measured by two axle-mount M-260 optical encoders (Encoder Products Co., Sandpoint, ID). These encoders were connected to a data acquisition system, LabJack U6 (LabJack Corp., Lakewood, CO) positioned under the seat and operating at a 400Hz sampling rate. A Windows tablet was used to store encoder data and provide power for the LabJack. Linear wheel velocity was calculated using **Equation 1**:

$$\nu = R\omega \qquad \textbf{(1)}$$

where ν (m/s) is the linear speed of DW, ω (radians/s) is angular speed, and R (m) is the radius of DW. The instrumentation added 2 kg to the wheelchair and is positioned nearby the CoM, such that it does not impact the system turning inertia. The setup of the measurement system is shown in Figure 4.

Wheelchair configurations. Aim #1 used the TiLite Aero Z (Kennewick, Washington) for all coast-down tests. The Aero Z had a mass of 12.1 kg and is coded as an ultralightweight manual wheelchair (K0005) by the Centers for Medicare and Medicaid Service [70]. The wheelchair was chosen based on its adjustability. The default wheelchair setup had 24" spoke wheels with 1 3/8" light gray tread pneumatic tires (Primo Orion) and 5×1 ½" urethane casters (EPIC) with aluminum forks (TiLite). The axle position was set at 7.6 cm forward of the backrest, resulting in 70% of wheelchair mass on its DWs. To evaluate different levels of resistive loss without changing system inertia, the wheelchair was configured at three DW tire inflations: 75 psi (100%), 55 psi (75%), and 38psi (50%).

Coast-down test

A coast-down test was performed to calculate deceleration parameters during free-wheeling straight and turning motions. According to our previous studies, wheelchair users typically maneuver at speeds < 1 m/s [3, 5] so speeds reflective of everyday mobility were selected for the coast-down procedure. A fixed-wheeled turn was achieved by engaging the wheel lock to one DW by a standard wheelchair lock. Both left and right turns were evaluated. Eight coast-down trials were conducted for each configuration by the same operator. The test protocol consisted of three phases: (1) push phase: An operator pushed the MWC to a speed of approximately 1 m/s during straight trajectories and 0.4 m/s during turning trajectories. These target speeds were chosen to ensure coast-down include the speed ranges for data analysis; (2) release phase: the MWC was released at the same location and direction within each trajectory; (3) free deceleration phase: DW velocities were measured as the wheelchair decelerated. Deceleration values were calculated over targeted speed ranges (straight: 0.65-0.95 m/s; turning: 0.1-0.3 m/s).

Protocol

Eight coast-down trials consisted of test repeatability and took place on separate days. To further validate the coast-down protocol, we evaluated the impact of tire inflation on energy loss along a straight and turning trajectories on tile.

Calculating the resistive forces that decelerate a rolling wheelchair requires the inertial properties of a loaded wheelchair, wheels, and casters. Mass, yaw inertia of the wheelchair (Izz), and the location of the center of the mass (CoM) were measured experimentally using iMachine. By measuring the CoM distance from the rear axle, we can calculate the caster-to-drive wheel weight distribution of the wheelchair.

Data Analysis

To capture CoM decelerations, time-series velocity data from both DWs were captured and low-pass filtered at 20 Hz. During the straight coast-down test, the averaged velocity values from left (V_{LD}) and right DW (V_{RD}) were used to represent CoM velocity (V_{CoM}). In turning trajectories, the tangential velocity ($\mathrm{V_T}$) of CoM was estimated based on the average of V_{LD} and V_{RD}. Deceleration during each trial was determined using linear regression. The coefficient of determination (r^2) in all linear regression models were > 0.96. **Figure 5** shows an example of velocity versus time during coast down.

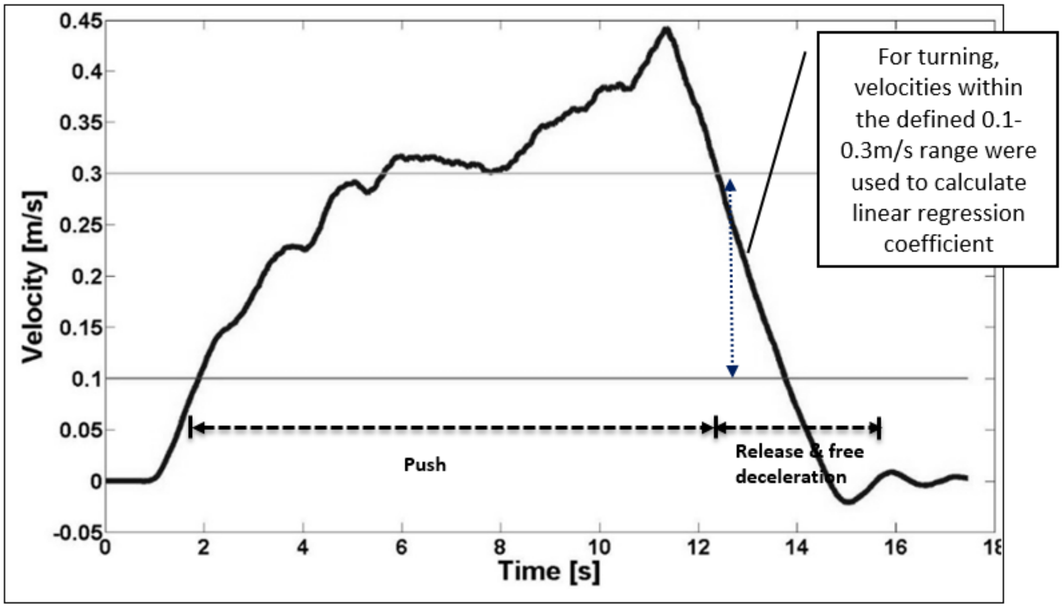


Figure 5 Example of a time versus velocity response during a fixed wheel turning coast-down test

Post-processing of all deceleration values was performed via a custom MATLAB code (MathWorks, USA). Repeated test variation is represented by the coefficient of variation (CV) and standard deviation (SD). To support future studies in coast-down tests, we analyzed the minimum number of repeated tests need to provide a reliable estimate of deceleration. The Spearman-Brown Prophecy Formula was used to predict the number of tests required to achieve high repeatability (alpha = 0.9).

To demonstrate test-retest reliability of the protocol, one operator repeated experiments on two different days. Reliability was characterized by the intraclass correlation coefficient (ICC) using a two-factor mixed effects model and type consistency. Since the coast-down testing sequence on Day 1 was not controlled to match the testing sequence on Day 2, deceleration values were randomized three times for each trajectory task before running the reliability test. The averaged ICC value was calculated for the final report.

2.1.2 Results

<u>Repeatability</u>

The results show that the coast-down protocol exhibits high repeatability with a low coefficient of variation (CVs <5.3%). Results of the Spearman-Brown Prophecy analysis suggest that three repeated trials are sufficient for achieving a reliability of 0.99 for all wheelchair setting. In addition, a high degree of reliability was found between two days by the same rater (ICC= 0.959). The mean between day variations for decelerations was 0.001±0.006 m/s^2.

<u>Validity</u>

By using the same mass and axle position, **Table 1** and **Figure 6** show the wheelchair decelerations with three different inflation levels. No differences in deceleration values existed across left and right turning trajectories (p=0.14) so averaged turning deceleration values are reported. Decelerations were greater during turning trajectories (Mean (M) = -0.122, standard deviation (SD) = 0.001 compared to straight trajectory (M = -0.073 SD= 0.001) averaged across tire pressures, p=0.001. The confidence intervals (Figure 6) and effect sizes (**Table 1)** indicate significant differences between all inflation levels. Effect sizes, represented as Cohen's d can be interpreted as the average percentile standing of the tested configuration relative to the 100% inflation level. As an example, the effect size of 2.7 places the mean of the 55psi configuration over the 99th percentile of the 75 psi torque distribution. Tires inflated to 75% had 10%

greater decelerations than 100% inflation in straight, and 14% greater decelerations in turning trajectories. In addition, 50% inflation had 18% greater decelerations than 100% inflation during straight trajectories, and 28% greater decelerations during turning trajectories.

Table 1 Deceleration values in each coast-down test by using test dummy with the default configuration

TP	Trajectory	Mean ±SD (m/s^2)	CV (%)	% increase	Cohen's d
75	Straight	-0.067±0.002	3.2%		
55	Straight	-0.074±0.003	4.0%	10.4	2.75
37	Straight	-0.079±0.002	2.0%	17.9	6.00
75	Turning	-0.107±0.005	5.3%		
55	Turning	-0.122±0.002	2.5%	14.0	3.94
37	Turning	-0.137±0.003	3.4%	28.0	7.28

TP: Tire pressure (psi); SD: Standard deviation; CV: Coefficient of variation

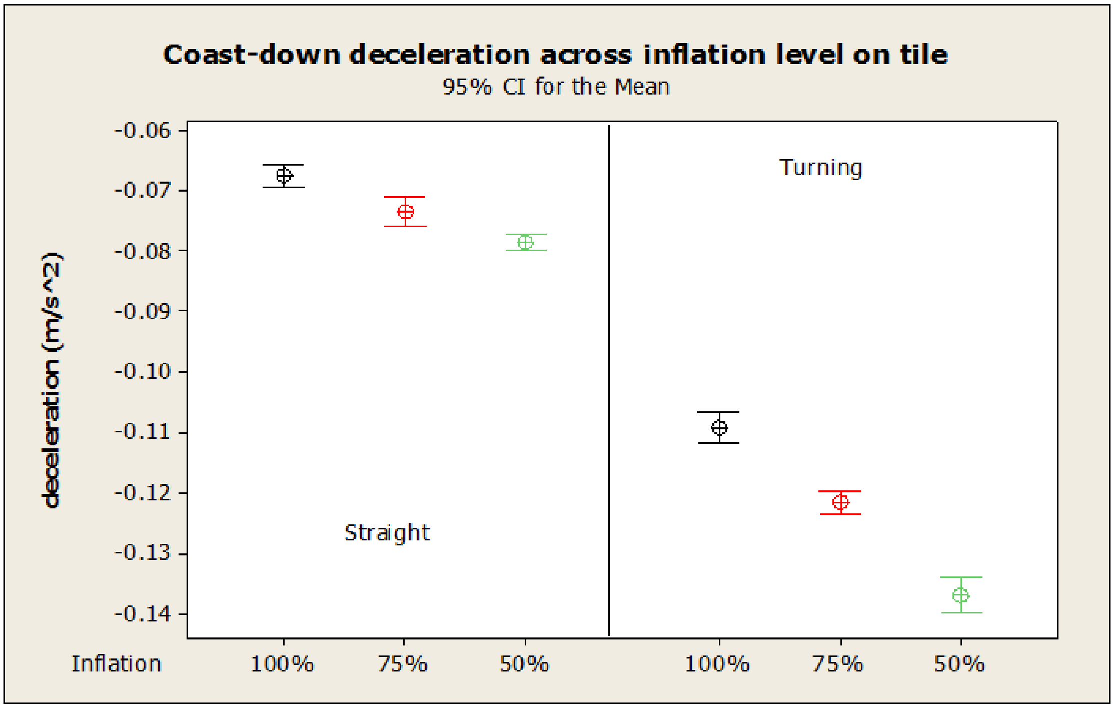


Figure 6 Comparison of deceleration for three tire inflations during straight and turning trajectories on the tile. CI: confident interval

2.1.3 Discussion

The described test method showed high repeatability (CV ≤ 5.3%) and reliability (ICC = 0.96) during both straight and fixed-wheel turn maneuvers. Analysis indicated that three trials would offer sufficient repeatability using the Spearman-Brown Prophecy formula. This repeatability is consistent with that of other free-wheeling coast-down tests. In Coutts's study [71], the decelerations during the first six trials were determined with an average CV of 7.1 % on a hardwood floor. Comparing the decelerations across studies is difficult since different coast-down studies used different wheelchair frames [32, 33, 62], wheels [31, 62], and ground type [16, 64, 72]. However, with similar weight distribution (30% loading on casters), our deceleration values had a similar range as Sauret et al. [62] study (from -0.05 m/s^2 to -0.25 m/s^2) on a hard smooth surface.

Resistive losses impact the overall propulsion effort of the user because they stem from forces that are always acting opposite to the wheelchair's direction of motion. These resistive forces and torques are ever-present, and in fact, are needed to maintain contact with the ground while maneuvering. The magnitudes of the forces and torques measured in this study were not very high with resistive forces being between 6-7 N and resistive torques between 3 ½ and 5 1/5 N-m. However, their cumulative effects over the course of the day can be significant. Research indicates that MWU travel about 1 km per day in about 90 bouts of mobility [3]. When maneuvering a chair with greater resistive losses, the user will have to exert greater effort within a bout of mobility and cumulatively, will expend more energy throughout the day. Because maneuvering wheelchairs include both straight and turning trajectories, a review of the respective resistive losses during both trajectories should be considered.

The fixed-wheel turn highlighted the resistive losses due to tire scrub. Results showed that the decelerations were significantly higher during turning compared to straight coast-down tests. Since resistive forces during turning include tire scrub and rolling resistance, these results illustrate that more resistive loss is associated with turning compared to propelling in straight directions. The results also highlighted the influence of tire inflation on resistive loss during turning compared to straight trajectories. With the increased contact area of a deflated tire, the results demonstrated that even a 25% decrease in tire inflation has an adverse effect on resistive losses. Moreover, the results indicated that inflation had a greater impact on resistive loss during scrubbing compared to rolling.

Prior studies have reported that tire type and pressure influence rolling resistance and thus can change propulsion effort [30, 31, 61]. Pneumatic tires have been shown to have lower rolling resistance than solid tires [31, 61]. Moreover, pneumatic tires with different inflation levels have been shown to have different rolling resistances [30]. By using the same type of pneumatic tires, Sawatzky et al. [30] found that a tire inflated to 100% had significantly less rolling resistance than tires inflated to 50%, but did not find a difference comparing tires inflated to 100% and 75%. Our results both corroborate and extend these results to add to knowledge about tire inflation. During straight trajectories, wheelchair deceleration at 75% and 50% inflation levels were 10% and 18% greater than that at 100% inflation, respectively. However, during turning, inflation levels had more influence, with deceleration increasing by 14% and 28% at 75% and 50% inflation levels, respectively. Significant differences in decelerations were found across all three inflation levels with large effect sizes. The results were also consistent with a previous study that indicated propelling a wheelchair through a 90-degree turn with one hand required 20% higher propulsion work than straight ahead on carpet [73]. Clinically, these results illustrated that turning effort will increase even with slightly deflated tires, thus emphasizing the importance of periodic wheelchair tire maintenance. With respect to measuring resistive losses, these results demonstrate that considering both straight and turning maneuvers can improve the sensitivity for differentiating resistive forces across wheelchair configurations.

2.2 Aim # 1.2: Identify the frictional energy loss across wheelchair configurations

2.2.1 Methods

Wheelchair configurations

Aim #1.2 was designed to evaluate the influence of mass and weight distribution on the resistive loss. For this purpose, the TiLite Aero Z wheelchair was re-configured to have a similar mass (17.6kg) and weight distribution (55% loading on DWs) as a K0001 standard wheelchair, the Invacare EX2 (Elyria, Ohio). The same coast-down protocols were conducted to measure the resistive losses of a wheelchair with 2 masses (12 kg vs. 17.6 kg) and 2 weight distributions (55% vs.70% loading on DWs). We chose to test the same wheelchair with different configurations rather than different wheelchairs in order to better control for potential confounding factors such as bearing resistance, tire type, and frame flexion. This approach allowed for the differences in results to be attributed to the independent variables, namely tire inflation, mass, and weight distribution.

Resistive losses: rolling resistance and tire scrub

The summation of resistive losses includes rolling resistance, tire scrub, bearing resistance, frame flexion, gravity, and variable external resistances. For our purposes, all motion was assumed to take place over level ground. Furthermore, given relatively slow speeds (less than 1 m/s) and use of the same wheelchair frame and wheels across trials, we assumed that the influence of aerodynamic drag and variable external resistance, such as bearing resistance could be negligible [74]. Figure 2 illustrates the simplified frictional forces that must be overcome in both straight and turning given these assumptions.

Drawing from the Sauret model for manual wheelchairs [62], the deceleration and associated resistive forces for a straight coast-down can be described as **Equation 2**:

$$\left(M_{sys} + 2\frac{I_d}{{R_d}^2} + 2\frac{I_c}{{R_c}^2}\right)\ddot{x} = -2\frac{\lambda_d}{R_d}(F_N)_d - 2\frac{\lambda_c}{R_c}(F_N)_c \qquad \textbf{(2)}$$

where M_{sys} is the system mass, $\ddot{x}$ is the tangential acceleration of CoM, I_d and I_c are the inertias of the DWs and casters, R_d and R_c are the radii of the DWs and casters, $(F_N)_d$ and $(F_N)_c$ are the normal forces on the DWs and casters, and λ_d and λ_c are the DWs and casters rolling resistance parameters (in meters), characterized by the fore-aft distance between the theoretical and actual centers of pressure in the wheel contact areas. Based on the small impact of the component inertias observed in Sauret's model simulations [62], it becomes possible to approximate the combined mass terms as the dominant system mass. Additionally, we can rewrite the normal force terms as rolling resistance, as shown in **Equation 3**:

$$M_{sys}\ddot{x} = -2(F_{RR})_d - 2(F_{RR})_c \qquad \textbf{(3)}$$

where $(F_{RR})_d$ and $(F_{RR})_c$ are the rolling resistance forces of the DWs and casters.

Compared to straight trajectories, resistive torque during fixed-wheeled turns includes not only rolling resistance but tire scrub [29, 34]. In the case of the rolling DWs and casters, tire scrub arises from a side-slip force that is generated due to a discontinuity between the tire heading direction and direction of velocity, otherwise known as the side-slip angle [63]. However, for the turning resistance model, the tire scrub associated the casters will be considered negligible. Since casters are linked to the wheelchair frame via vertical pin joints, the transmitted torques under ideal conditions are zero. Furthermore, during a turning coast-down test, the casters have become aligned with the radius of curvature, reducing the turning resistance that arises from the side-slip forces transmitted through the forks. The scrub torque of the fixed and pivoting DW is fundamentally different and is greater in magnitude than that of the casters and the rotating DW since the same contact patch is constantly undergoing shear. For example, the coast-down forces during a fixed-wheel left turn can be modeled as **Equation 4**:

$$\mathrm{I}_{turn}\left(\frac{\ddot{x}}{r_{G/O}}\right) = (r_{A/O}) \times -(F_{RR})_A + (r_{B/O}) \times -(F_{RR})_B + (r_{C/O}) \times -(F_{RR})_C - (\tau_{ss})_C - (\tau_{ss})_O \quad \textbf{(4)}$$

where I_{turn} is the combined inertial terms (kg-m^2) of the system and its components during fixed-wheel turning; $\ddot{x}$ is the tangential acceleration of the CoM; $r_{G/O}$, $r_{A/O}$, $r_{B/O}$, and $r_{C/O}$ are the distances from the center of rotation to the CoM, left caster fork, right caster fork, and the right DW; $(F_{RR})_A$, $(F_{RR})_B$, and $(F_{RR})_C$ are the rolling resistance forces of the left caster, right caster, and right DW; $(\tau_{ss})_C$ and $(\tau_{ss})_O$ are the scrub torques of the rolling right DW and the fixed left DW. It should be noted that the I_{turn} term is heavily dominated by the system yaw inertia, I_G, adjusted by the parallel-axis theorem to be centered at point O (the center of rotation) and represented by I_O, as shown in **Equation 5**.

$$\mathrm{I}_O = \mathrm{I}_G + M_{sys}\left(r_{G/O}\right)^2 \quad \textbf{(5)}$$

By applying the same simplifying assumption for Equation 3 that allows us to treat component rotational inertias as being negligible, Equation 4 can be reduced as shown in **Equation 6**.

$$\mathrm{I}_O\left(\frac{\ddot{x}}{r_{G/O}}\right) = (r_{A/O}) \times -(F_{RR})_A + (r_{B/O}) \times -(F_{RR})_B + (r_{C/O}) \times -(F_{RR})_C - (\tau_{ss})_C - (\tau_{ss})_O \quad \textbf{(6)}$$

Generally, overall resistive loss during straight trajectories is mainly contributed by wheel rolling resistance (**Equation 3**), whereas both rolling resistance and tire scrub contribute to overall resistive loss during fixed-wheel turns (**Equation 6**). Therefore, tire side-slip force and its resultant scrub torque is the distinguishing resistive loss factor between these two maneuvers.

Statistics

Decelerations for the chairs with different masses and weight distributions during both trajectories were tabulated and graphed using means and confidence intervals. Overall resistive force in a straight motion and resistive torque in turning were later calculated based on the measured decelerations as well as respective system mass or inertia. This data provided the most direct evaluation of differences in the ability to judge meaningfulness across wheelchair designs.

The test sensitivity was assessed by its ability to distinguish differences in resistive losses across different MWC configurations, specifically DW inflation levels, masses, and weight distributions. In a strict sense, testing multiple wheelchair configurations over multiple trials does not permit the use of ANOVA to infer differences due of the violation of the assumption of independence. However, in deference to convention, simple univariate ANOVA results are reported for the straight and turning maneuvers using $p<0.05$ to define statistical significance. Percent differences and effect sizes were calculated to describe fully differences in decelerations across configurations.

2.2.2 Results

Table 2, **Table 3**, and **Figure 7** show the wheelchair decelerations, related resistive forces and torques with 2 masses * 2 weight distributions. Analysis of deceleration values indicated that values across both mass and weight distribution were significantly different for both trajectories (*p*<0.001). In straight trajectories, wheelchairs with a 17.6kg mass had, on average, 2% greater decelerations compared to those with 12.1 kg mass and wheelchairs with 55% weight on their DWs had, on average, 17% higher decelerations. In turning trajectories, wheelchairs with a greater mass had 11% greater decelerations, while wheelchairs with 70% weight of their DWs had 32% greater decelerations. Differences between the base configuration of the ultralightweight wheelchair (12.1 kg & 70%) and the other configurations are indicated by percent differences and Cohen's d effect sizes (**Table 3**). All effect sizes can be considered as large [75]. As a reference, the effect size of 0.63 places the mean of the 17.6 kg & 70% configuration value at the 74th percentile of the reference configuration- in this case, 12.1 kg & 70% configuration.

Table 2 Wheelchair decelerations across 2 masses * 2 weight distributions during straight and turning trajectories

	Straight trajectories			
Configuration	**Decel. [m/s^2]**	**% difference**	**Effect size**	
12.1kg & 70%	-0.067±0.002			
17.6kg & 70%	-0.068±0.001	1.5	0.63	
12.1kg & 55%	-0.078±0.002	16.4	5.50	
17.6kg & 55%	-0.080±0.002	19.4	6.50	
	Turning trajectories			
Configuration	**$r_{G/O}$ [m]**	**Decel. [m/s^2]**	**% difference**	**Effect size**
12.1kg & 70%	0.287	-0.109±0.005		
17.6kg & 70%	0.287	-0.119±0.006	9.2	1.81
12.1kg & 55%	0.328	-0.081±0.005	-25.7	-5.60
17.6kg & 55%	0.328	-0.092±0.007	-15.6	-2.80

Decel.: Deceleration; $\mathbf{r_{G/O}}$: the radius between the CoM and rotation point.

Table 3 The resistive forces and torques across configurations during straight and turning

	Straight trajectories	Turning trajectories
Configuration	**Resistive force [N]**	**Resistive torque [N·m]**
12.1kg & 70%	-6.096±0.181	-4.715±0.216
17.6kg & 70%	-6.518±0.096	-5.338±0.269
12.1kg & 55%	-7.083±0.181	-3.530±0.219
17.6kg & 55%	-7.666±0.193	-4.210±0.320

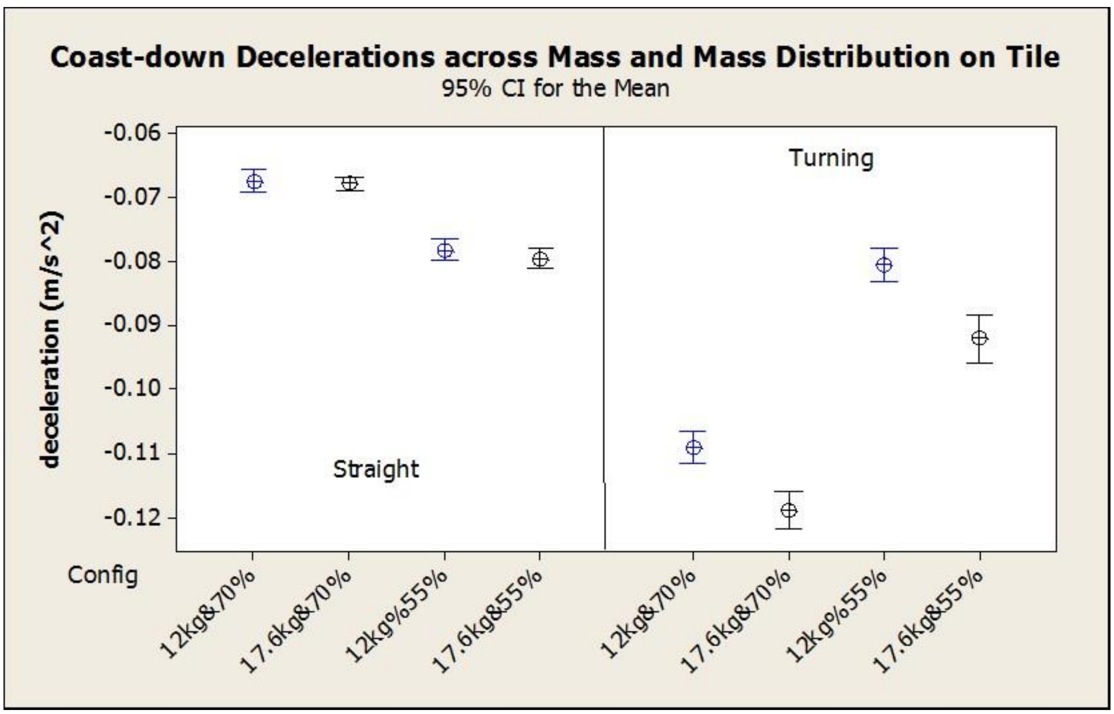


Figure 7 Comparison of deceleration for four configurations during straight and turning trajectories on the tile. Confi: configuration; i.e., 12kg&70%= 12kg wheelchair with 70% loading on DWs

2.2.3 Discussion

The results indicate that weight distribution had a greater influence on resistive forces than mass in both straight and turning. In addition, the effect of weight distribution on resistive loss varied according to the trajectory of motion. This result is consistent with those of Bascou et al. [33] who studied rolling resistance of different wheelchair configurations, and reported a 52% increase in the decelerations when the loads on the casters varied from 29% to 64% using the same wheelchair mass [33]. Similar observation was made in a study by Sauret et al. [62], in which they speculated that decelerations would continually increase as the distribution of total mass on the front wheels increased. These studies underlined the significant influence of weight distribution on the resistive force in straight trajectories, and our results extend these findings to describe the impact of weight distribution on resistive torque in turning. According to our results, the frictional force of the chairs with 55% weight on the DWs was 17% greater than those with 70% weight distribution in a straight trajectory. In distinction, the chairs with a 70% weight distribution had resistive torques that were 30% greater than those with a 55% weight distribution during turning. This is an important finding that can be explained by the different sources of resistive loss in straight and turning trajectories. When traveling straight, resistive forces are due to the rolling resistance of the casters and DWs. The results show that a greater weight distribution on the DWs reduces the system’s rolling resistance. In a fixed-wheel turn, resistive torques are due to rolling resistance of the casters and the rolling DWs as well as the losses due to tire scrub. Tire scrub happens when a wheel rotates on the rolling surface during a turning maneuver. The results showed that the greater load on the DWs due to moving the axle forward results in a greater resistive loss because of the heightened tire scrub.

Current clinical practice advocates for a forward axle position based on its advantage from a biomechanical perspective [36, 37]. Pitch stability is rightly identified as an important consideration when determining how anterior the axle should be positioned. Pitch stability decreases because a forward axle position shifts the center of mass rearward with a concomitant increase in weight on the drive wheels. However, our results demonstrate the increase in biomechanical advantage also comes at the cost of increased turning resistance. Like many wheelchair configuration changes, axle position impacts multiple performance variables both negatively and positively. As a result, clinicians should consider this tradeoff when adjusting wheelchairs. In particular, propulsion effort should be evaluated as wheelchair users perform turning maneuvers on common surfaces, including tile and carpet.

2.2.4 Conclusion

The dominant resistive forces of a rolling wheelchair result from rolling resistance and tire scrub. Multiple researchers have measured coast-down in straight trajectories, yet, turning maneuvers must also be considered to characterize the resistive loss. The described coast-down test method, including straight and fixed-wheel turns, offers a simple and reliable method for assessing manual wheelchair resistive loss. This method could easily be applied to evaluate the influence of different wheelchair configurations, tires, and rolling surfaces. The results of this study confirmed that higher resistive loss exists during 1) turning compared with straight trajectories, and 2) under-inflated pneumatic tires. In addition, the results indicate that weight distribution has a greater impact on resistive losses compared to a 5.5 kg increase in wheelchair mass and that this impact varies between straight and turning maneuvers. Based on the results of this and previous studies, clinician and wheelchair users should carefully consider the impact of wheelchair resistive loss while selecting or configuring a manual wheelchair. Researchers should also consider how differences in resistive losses during turning maneuvers impacts propulsion effort. Based upon prior research, a biomechanical benefit exists with a forward axle position. However, the concomitant increased weight on the DWs can increase resistive torques during turning. Additional research is needed to define axle positions and weight distributions that optimize propulsion efforts during turning trajectories by balancing the biomechanical benefit against the increased resistive loss due to tire scrub.

CHAPTER 3
TO DEVELOP AND VALIDATE A TESTING PROTOCOL FOR MEASURING MAXIMUM PROPULSION STRENGTH

3.1 Aim # 2.1: Develop a testing procedure for measuring maximum isometric propulsion strength while subjects are seated in their wheelchairs

Studying muscle strength for wheelchair users is always an essential topic since it has been shown to influence the incidence of chronic shoulder pain [76] and wheelchair propulsion [43, 47, 77]. During manual wheelchair propulsion, various kinetic [47], electromyography [78, 79], and musculoskeletal modeling [80] studies have confirmed that shoulder and elbow muscle are the key muscles to generate the propulsion force. These muscle groups further increase the demands when accelerating the wheelchair from a complete resting position [36, 39] or increasing speed [81].

Regarding muscle strength of upper extremity, shoulder adductors and handgrip strength was shown to be associated with improving propulsion speed significantly [77]. Shoulder flexion and elbow pronation also had more than 50% muscle demand during the propulsion cycle relative to its muscle capacity [47]. These muscle studies highlighted the importance to study muscle strength for wheelchair propulsion. However, limited studies have evaluated muscle strength directly while propelling the wheelchair. In addition, studying the strength of specific muscle group has limitations to fully representing the muscle performance related to propulsion task in a systematic way.

Propulsion strength, like other strength measures, is dependent on the task and environment. Wheelchair propulsion is a complex task, which is influenced by the upper extremity position [82] and hand interface [83] as influenced by of the wheelchair seat (i.e. person's arm angle relative to push-rim), push-rim design, and sitting posture [52]. As a result, changing either one of the mechanical or biomechanical parameters would influence propulsion strength outcome. The objective of this Aim was to develop and validate an approach to measure propulsion strength as a person is seated in his or her wheelchair.

Several design criteria were defined, including 1) to the ability to measure propulsion strength without changing users' wheelchairs settings, 2) compatibility with most commercially available wheelchairs, which includes different wheelchair frames and sizes, 3) being able to measure propulsion strength without requiring a transfer or changing seating postures, and 4) not impact the biomechanical relationship to the push rims. Finally, according to literature reviews in measuring muscle strength for people with spinal cord injury [43, 47, 84], the system should be able to handle the pushing force at least 310 N from each side.

3.1.1 Design selection

The final design selection of measuring propulsion strength is illustrated in **Figure 8**. The system includes two primary functions: 1) securing the wheelchair during measurement and 2) attaching force transducers to the drive wheels.

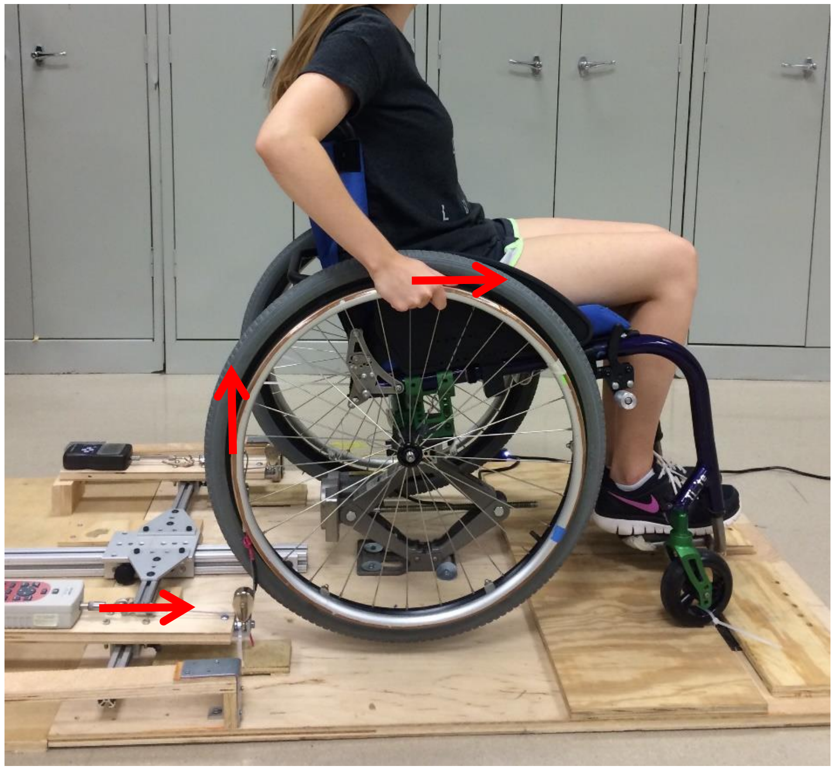


Figure 8 Side view of strength measurement design

The devices are all attached to a sheet of solid plywood, which serves a platform for the devices and the subject while sitting their wheelchairs. **Figure 9** is the overall design and placement of each part of the apparatus.

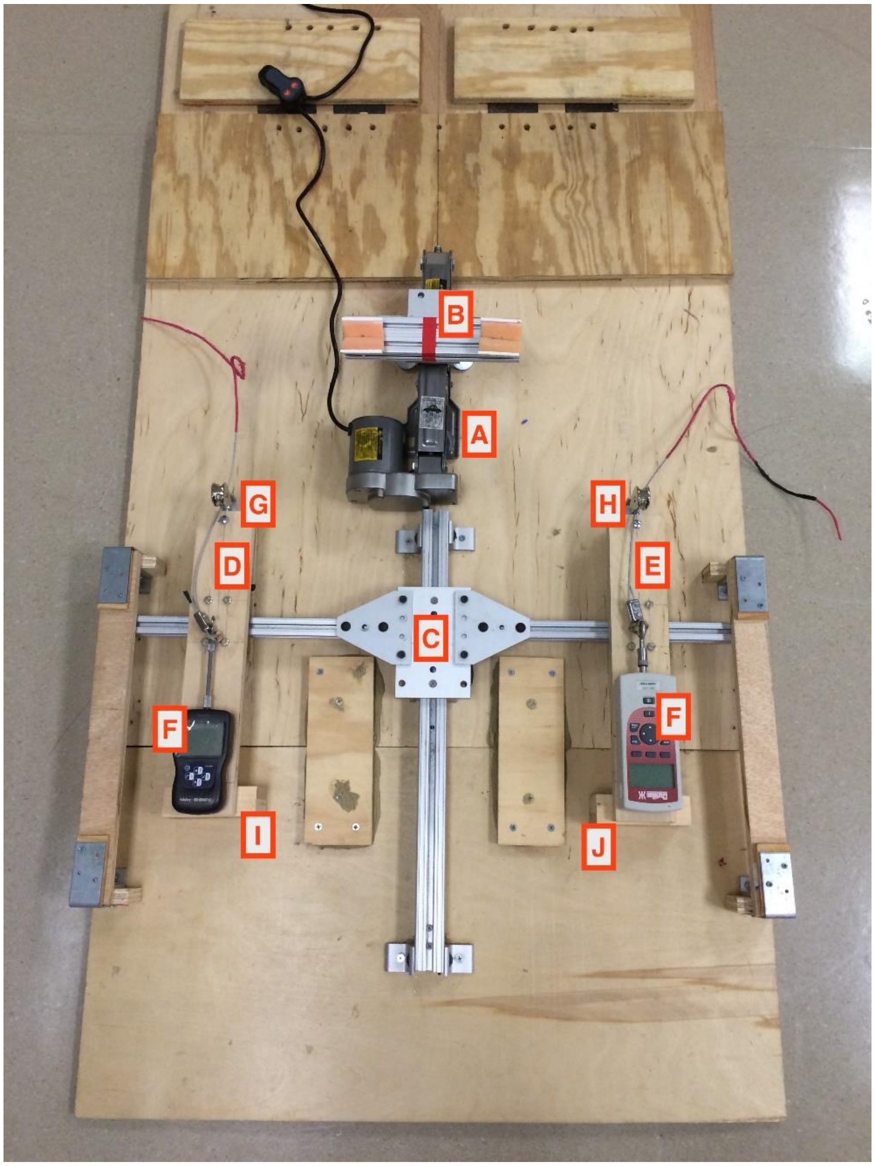


Figure 9 Overall design of wooden platform/placement of devices

The front half of the platform is where the subject's wheelchair placed. An electric lift (Figure 9 - A) with axle resting attachment (Figure 9 - B) is secured to the platform as is used to slightly elevate the wheelchair's drive wheels. This is required to permit measurement of the propulsion force applied to the wheel. The lift interfaces with the wheelchair frame using an attachment designed for either rigid (**Figure 10, A**) to folding (**Figure 10, B**) frames. The electric lift is attached to an electrical power source (12V, 6amp).

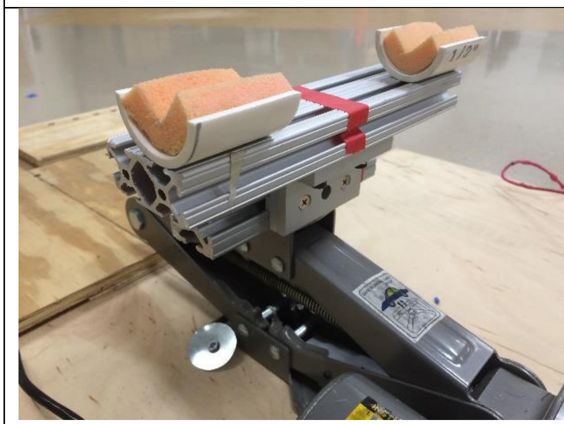
A	B
	

Figure 10 Rigid wheelchair axle resting attachment placed on electrical lift

On the back half of the platform, an adjustable cross arm (Figure 9 – C) is secured to the platform. This cross arm may be slid forward or backward on the platform based on how much space is taken up by the wheelchair drive wheels. The cross arm has two adjustable side arms (Figure 9 – D and E) on the left and right sides. Two force gauges, Shimpo, Electromatic Equip't Co., USA and Chaillon, Ametek, USA, are used to measure the maximum force exerted on each side (Figure 9 – F). The force measurement from both manufacturers is consistent (R^2=1) and reliable. Each force gauge is attached to a pulley (Figure 9 – G and H) that is also attached to that sidearm. Each pulley is aligned with either the left or right drive wheel axle while setting the wheelchair position. Both the pulleys hold a metal wire that is attached to the force gauge on the corresponding side arm. The wire is later attached to a position on the wheelchair's drive wheel that is horizontally aligned with the center of the axle of the drive wheel. A small wooden block

(Figure 9 – I and J) is placed underneath the back of each side arm, under the force gauge, to stabilize the side arm when the subject applies maximum force to the system. This configuration holds the wheel static when the occupant propels forward thereby permitting measurement of isometric force.

3.1.2 Method

Testing protocol. By rolling the wheelchair backward onto the platform, the researchers align the wheelchair axle with the axle resting attachment on the electrical lift. The attachment will later hold the wheelchair axle for lifting the wheelchair up. Wheelchair's two front caster wheels and wheelchair axle are secured by using plastic zip ties (**Figure 11** and **Figure 12 - A**). This step is to provide stability when the subject begins the test.

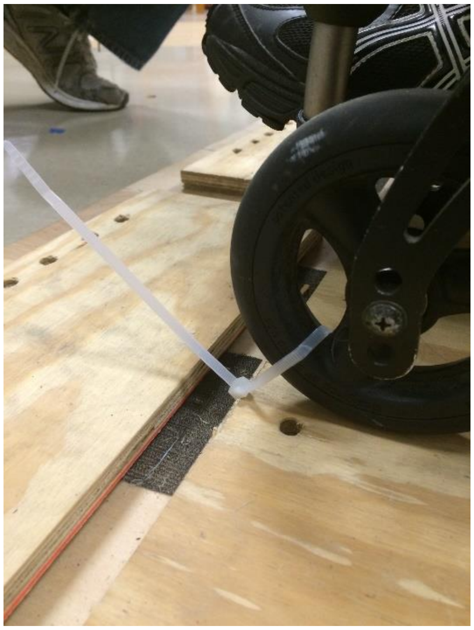


Figure 11 Zip-tie attach wheel to the wooden platform

The wheelchair frame is secured on a base with locked casters. The drive wheels are raised 1 cm off the ground, allowing subjects to push the wheels freely while not impacting the wheelchairs seat angle. The researchers adjust the cross arm (**Figure 9 – D**) so that the two pulleys on each side arm are directly behind the push rim on the drive wheels (**Figure 12 - B**).

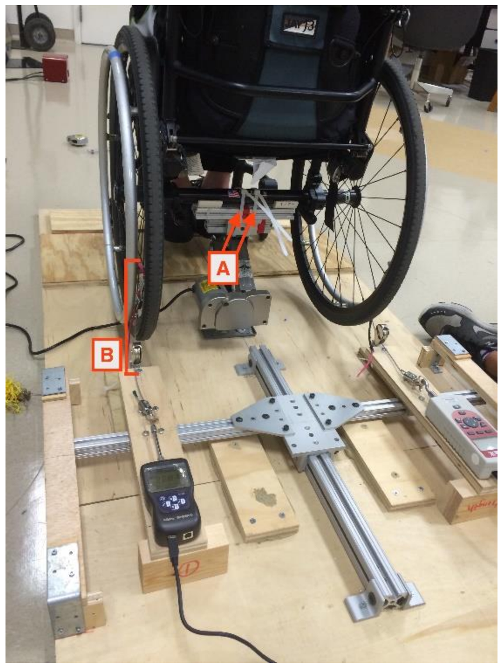


Figure 12 Rear view of wheelchair on electrical lift, showing overall placement of wheelchair

A pulley system attached to a force gauge behind the wheelchair apply static backward force to the wheels. A cable connected the wheelchair spoke and force gauge with a pulley set at 90 degrees. The cable setup is used to transfer the force from the top-dead-center (TDC) of each wheel to the force gauge.

The study records the force reading from both wheels separately. The test consists of 5 seconds maximum isometric force exertions of each arm from the TDC, a position selected based on the largest moment generated during propulsion [80]. We repeated this test four times with a two-minute rest between trials. The maximum values of the four repeated trials represent the maximum isometric strength. Overall, this study design meets design criteria by adapting to different wheelchair configurations without changing wheelchair setting and seating postures.

3.2 Aim # 2.2: Evaluate the repeatability and validity of the test protocol to measure maximum propulsion strength

The previous research considers muscle strength as one of the factors that limit the mobility of wheelchair users [40]. However, the testing of muscle performance related to wheelchair propulsion is still under development. Traditional approaches for measuring the muscular strength of disabled subjects include hand dynamometry [77], repetitive weight-lifting [85], and computerized isokinetic dynamometers (e.g. Cybex and Kin-Com) [47, 76]. It is widely agreed upon using measures of static strength for field-testing. However, these types of approaches lack a direct connection between muscle strength and wheelchair propulsion.

The purpose of Aim# 2.2 is to assess the repeatability and validity of the propulsion strength measurement system (Aim #2. 1). Repeatability was determined using repeated measurements of subjects over time and concurrent/predictive validity was determined using a comparison to isometric strength as measured using a Kin-Com. This maximum force may be comparable to the force required to propel their wheelchairs on cement or carpet surfaces. Having a reliable and valid quantitative measurement are fundamental to developing a quality approach in evaluating propulsion strength. By definition, the reliability of the protocol indicates that the measurements are consistent with repeated administrations of the test [86]. The validity, especially the concurrent validity, of the protocol indicates that the outcome of the experiment test can be used as a substitute measure for an established reference standard criterion test [86].

3.2.1 Methods

Subject. We recruited a convenience sample of healthy participants. Potential participants were excluded if they reported they had shoulder dysfunction that influences arm movements. The population was chosen in an attempt to have a homogeneous sample with a wide range of muscle strength and body size. The physical characteristics of participants were described in **Table 4**. This cohort was selected because the Kin-Com protocol for shoulder flexion requires test subjects to be seated in the device and secured in a specific manner. This procedure has not been validated for wheelchair users as it requires a very complex transfer and is not designed to provide consistent trunk support that may be needed to isolate shoulder flexion force.

Propulsion strength measurements. The detail testing procedure was described in Aim # 2.1. For testing with healthy subjects, wheelchairs were selected randomly and adjusted to match their body size (e.g. seat width, depth, and height). Subjects were instructed to sit upright comfortably, grab the top dead center of the wheelchair push-rim, and push the push-rim forward as hard as they can. The pushing consisted of 5s maximal isometric force exertions of both arms together with four repeated trials.

Kin-Com isokinetic dynamometer. During wheelchair propulsion, peak wheelchair propulsion strength in shoulder flexion can account for up to 67% of maximum isolated shoulder flexion strength [47], which means the largest joint moment is during shoulder flexion. Therefore, to validate our method of measuring maximum isometric propulsion force, we used the Kin-Com isokinetic dynamometer following its validated protocol for shoulder flexion.

The participants were tested in a secured and standardized seated position. The right shoulder was tested in the following position: 45° flexion and neutral rotation, 15° abduction, 90° elbow flexion, and neutral forearm rotation [47, 76]. The axis of rotation of the Kin-Com level arm was also aligned with the flexion-extension axis of the shoulder. To compare the propulsion force, we have chosen the right side of maximum isometric contraction—the same contraction type as that of the technique used in this study.

Protocol. The same examiner conducted the test-retest reliability protocol by doing the same task on two different days. The same examiner also conducted the validity test by comparing in-chair isometric propulsion strength to a standardized test of shoulder strength using a Kin-Com dynamometer. During the test, four trials were repeated, where in each trial, the maximum force was measured when the subject propels his or her drive wheels.

Statistics. Descriptive statistics were used to analyze the strength values and the subject characteristics. Intraclass correlation coefficients (ICC) with a two-factor mixed effects model and type consistency [87, 88] were used to assess the test-retest reliability. ICC (2,1) for intrarater reliability were calculated by comparing all measurements in different days following the same protocol. To evaluate the predictive validity, the non-parametric Spearman's rho rank-order correlation (γ_S) was computed to establish the relationship between propulsion and Kin-Com strength measurements. The maximum propulsion strength measurements from both sides were averaged, whereas only the right side of Kin-Com strength measurements was used. The value was later fed into the statistical software, SPSS 22.0 for computing. The significance level was defined as $p < 0.05$.

3.2.2 Results

Eight healthy subjects (female:2; male:6) were recruited to participate in a test-retest reliability test and ten healthy subjects (female:4; male:6) were recruited to participate in a validity test. All the subjects in this study reported right dominance (right-handed and used preferentially the right upper limb in daily activity). Table 4 shows the description of subjects' characteristics.

Table 4 Physical characteristics of recruited subject

	Age	Height (cm)	Body weight (kg)	Propulsion strength (N)	Propulsion strength (N/kg)
Reliability (n=8)	27±4	172.1±8.4 (157-182)	70.9±14.7 (48.5-88.5)	Day 1: 207.3±53.2 (150.2-307.2) Day 2: 193.9±55.8 (132.6-299.4)	Day 1: 2.7±0.5 (2.1-3.4) Day 2: 2.5±0.7 (1.7-3.6)
Validity (n=10)	27± 6	170.0± 8.2 (157-182)	67.7±13.3 (47-84)	186.3±39.9 (118.0-252.0)	2.7±0.5 (2.1-3.4)

<u>Reliability</u>

The study showed a high degree of reliability between propulsion strength measurements in two different days. The ICC was 0.973 with a 95% confidence interval from 0.866 to 0.989. The mean difference in measuring maximum propulsion strength between days was 13.4 ± 6.7 N. **Figure 13** demonstrated the relationship between day 1 and day 2 measurements.

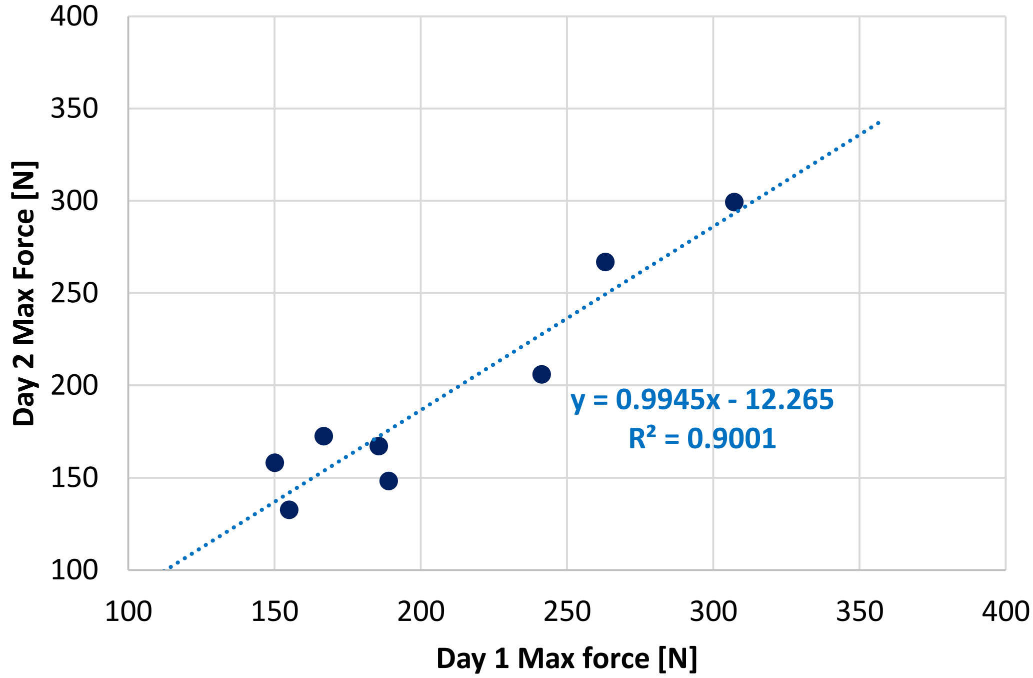


Figure 13. Averaged left and right maximum propulsion strength

Validity

The validity test was performed statically through comparing the propulsion strength to shoulder flexion strength in the right side. The **Figure 14** illustrates the result of experimental measurements, gold standards, and reflected R-square value. The results indicated a high linearity (Spearman's rho, γ_s=0. 815, *p*<0.01) between the experimental measurements and gold standards.

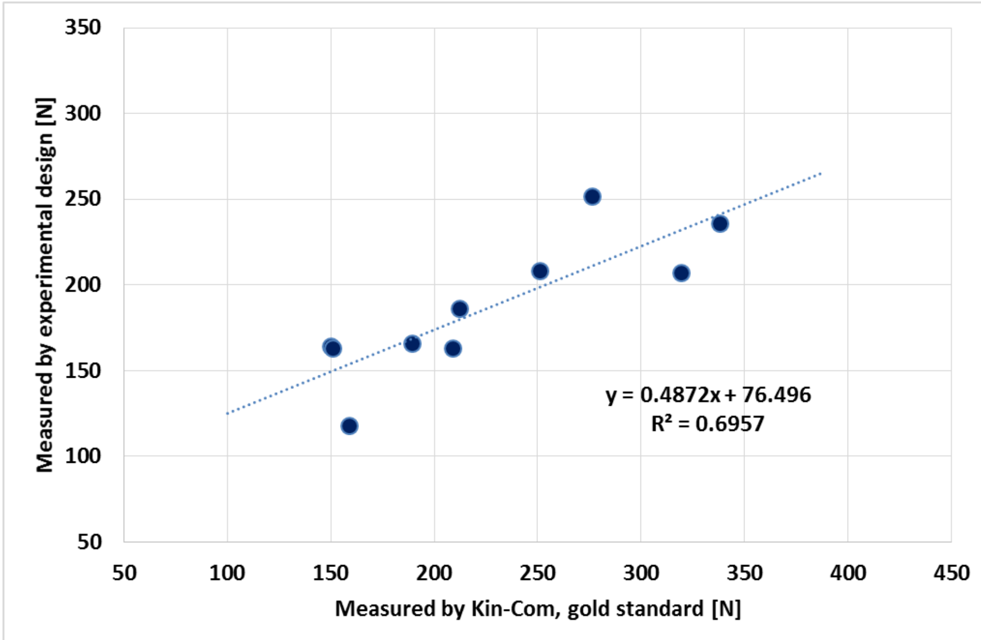


Figure 14. Strength comparison between gold standard and experimental design

3.2.3 Discussion

The purpose of this study is to implement and evaluate a new approach to measuring propulsion strength without changing wheelchair configurations and seating. Compared to previous approaches in measuring muscle strength, this experimental design had the features to accommodate different wheelchair configurations and seating without difficulty. As the main finding, the experiment protocol demonstrated a reliability and validity similar to those reported for other measures of isometric strength.

Our technique had an acceptable reliability with ICC=0.973. A recent systematic review of upper extremity dynamometry recommended using measurements with an ICC > 0.9 [89] while acknowledging that some measurements tend to report lower values. For example, the hand-held dynamometer is one of the common techniques used to evaluate isometric strength in different positions and populations [90]. According to May's et al.

study [91] in assessing the reliability of the hand-held dynamometer (Penny and Giles Transducer Myometer, England) for measuring shoulder rotation strength, the ICC of the intrarater reliability ranging from 0.89 to 0.96. Other studies using Nicholas Manual Muscle Tester (MMT) showed a high intrarater reliability in measuring shoulder [92, 93] and elbow [90, 93] strengths with high ICC values ranging from 0.84 to 0.97 and from 0.80 to 0.99, respectively. The study using another type of hand-held dynamometer (Chatillon CSD400C) also show a high test-retest reliability in measuring shoulder, elbow, wrist, hip, knee, and ankle strength with high ICC values ranging from 0.93 to 0.98 [94]. Compared to previous strength measurement studies, our approach also provided a high reliability with ICC = 0.973.

Regarding the validity of propulsion strength measurements, our values were also reasonable and highly correlated with dynamometer measurements (Spearman's rho, γ_s=0. 815, p<0.01). Previous studies have shown a high correlation between hand-held and isokinetic dynamometer on measuring shoulder [91], elbow [90], and knee [90, 95] isometric strength with Pearson γ value 0.86 to 0.88, 0.74, and 0.57 to 0.80, respectively. We anticipated a difference exist between the propulsion and dynamometer isometric strength measurements. In our case, propulsion strength measurements were lower than the dynamometer measurements. The reason may be because of different testing positions, different directions of pushing force [49], techniques, and active muscles. For example, shoulder strength only explained around 60% of the variance of propulsion tasks [47, 77]. Other factors, such as handgrip strength [77], wrist strength [47], and trunk stability [77] could also influence the isometric propulsion strength measurements. In addition, the bilateral deficit can cause a difference in strength measurement. In our study, the maximum propulsion strength is measured during simultaneous bilateral muscle contraction. Therefore, the right side of propulsion strength could have a lower value than the force generated unilaterally due to neural inhibition [96].

Although the strength values were different, our propulsion strength results were close to another similar study. Compared to Janssen's et al. study [43], which built a device for the analysis of muscle strength in a stationary wheelchair ergometer, our design presented a similar propulsion strength measurement. By normalizing body weight, our strength measurements from healthy subjects ranging from 2.1 to 3.4 N/kg, which fell within the similar range (from 0.8 to 4.2 N/kg) as Janssen's et al. study with the SCI population.

3.2.4 Conclusion

We have developed and validated an approach that evaluates upper limb muscle strength related to wheelchair propulsion. Although we demonstrated the feasibility of maximum propulsion strength acquisition in healthy human subjects, our results are relevant since we included the variability of gender, age, body weight, and muscle strength and excluded the confounding factors, such as propulsion skills and injuries. This study also provides the quantitative measurements necessary for wheelchair configurations and seating. In our subsequent study, we used this technique as one of the physiological measurements to predict wheelchair propulsion effort.

CHAPTER 4
TO DEVELOP A REGRESSION MODEL THAT IDENTIFIES THE IMPACT OF USERS' PHYSICAL FACTORS AND WHEELCHAIRS' MECHANICAL PROPERTIES ON PROPULSION EFFORTS DURING OVER-GROUND MANEUVERS

4.1 Aim #3. 1: To develop and evaluate a repeatable propulsion course that highlights changes in speed and direction

Studying steady-state propulsion on treadmills, wheelchair ergometers, or straight tracks is common for many biomechanics and exercise studies. However, a short distance but slow propulsion speed dominates wheelchair usage in a natural environment [3, 97]. Therefore, to reflect "real life" wheelchair utilization, studying the performance of wheelchair propulsion should consider the pattern of initiating movement, turning, and stopping the wheelchair.

The propulsion course consisted of activities of daily living (ADL) has been used to evaluate driving comfort [98] and physiological response [99]. In DiGiovine's et al. [98] study, they included tile, carpet, and dimple strip. They also used a 50mm high 5° ramp with curb, and a 25mm, 50mm, and 75mm high-speed bump. Hayes et al. [99] also used a series of ADL courses including desk work, loading and unloading a dishwasher, transferring to and from a wheelchair at a self-selected pace, and performing laundry tasks. These ADL courses were well designed to represent common driving tasks encountered by users in their daily lives. However, these course designs are complex and limited in providing a consistent propulsion pattern, reaching moderate to high-intensity activity, and maintaining steady-state exercise conditions.

To emphasize the turning feature in wheelchair mobility, Mattison et al. [97] developed a realistic course with tortuous circuits, which included a large oval and large and small figures of eight. Reid et al. [100] also developed a propulsion course with various steering. This type of course design was reproducible and valid to show the difference in metabolic costs between propulsion methods [97]. By evaluating the impact of steering on metabolic costs, Reid et al. [100] further showed that subjects need a higher metabolic demand while propelling on tracks with sharp turns than with wide turns or on a treadmill. Although these study designs provided a simple circuit to measure overall wheelchair performance and highlighted the influence of steering on metabolic costs, these studies did not provide a sensitive measurement to quantify the impact of wheelchair configurations on propulsion efforts.

Designing a repeatable maneuver endowed with representative acceleration, stops, and turns to evaluating propulsion effort is challenging. Previous studies provided an alternative approach to evaluating wheelchair performance by combining either different basic ADL skills [98, 99] or wheelchair maneuvering [97, 100-105]. However, these studies did not demonstrate a repeatable maneuver that could study the impact of wheelchair design on propulsion efforts. Therefore, Aim #3. 1 is to develop a repeatable propulsion course that highlights changes in speed and direction. In addition, the protocol should be able to distinguish the difference in propulsion efforts across wheelchair designs.

4.1.1 Methods

Design Selection. To develop a valid and reliable approach to assessing propulsion effort, our over-ground maneuvers should 1) highlight the representative acceleration, deceleration, and turns in daily activities, and 2) last at least five minutes to reach steady-state VO_2 [106]. In addition, to improve repeatability by reducing fatigue and variance, the task should 3) be simple, 4) include reciprocal arm movement, and 5) control propulsion speed.

To highlight straight and turning trajectories under a long bout of activity, we investigated several propulsion courses. The propulsion courses including modified figure-8 with 540° turn course, modified figure-8 with 180° turn course, slalom course, one push, stop, and turn course, and two pushes, stop, and turn course.

Partitioning Kinetic Energy. The study used kinetic energy (KE) as an objective measurement to characterize the features of changing directions within propulsion course [107]. The detail of method and a mathematical model were described in Medola's et al. paper [107].

A wheelchair is viewed as an assembly of 7 rigid bodies: frame, left and right drive wheels (DWs), left and right caster forks, and left and right caster wheels. Mass and yaw inertia of the system, $I_{ZZ,sys}^{G}$, is measured experimentally using a device called the iMachine [67] that was developed for this purpose. Rotational inertia of the DWs, casters, and caster forks are measured using a system based on the established Trifilar Pendulum [68, 108], which measures the rotational inertia based on the mass and the frequency of oscillation.

The total energy of a wheelchair in motion is comprised of the sum of the kinetic and potential energy, whereas potential energy effects being neglected when the motion is on flat ground. Maneuvering the wheelchair over-ground, or freewheeling, requires force input to both DWs to impart KE. KE can be calculated using the inertias and motions of all seven components. Analysis keys off of the yaw rotation rate and velocity of the center of mass (CoM). DW rotation rates are measured using encoders mounted on each wheel. By measuring the DWs' rotation rates, one can determine the translational and rotational velocities of all other components.

The KE of the wheelchair in freewheeling on flat ground is the summation of the KE of its parts (Equation 1). During freewheeling motion, the KE terms can be partitioned into three broad categories: translational KE, rotational KE, and turning KE. Translational energy, the first term in Equation 7, represents the energy of linear motion. Rotational energy represents the energy of all four wheels spinning on their axes. Finally, turning energy is embodied by the terms containing the yaw rate $\dot{\Psi}$, which includes the yaw rates of the whole chair and the caster fork assembly. A full accounting of propulsion effort must be done using maneuvers endowed with all three types of KE.

$$KE = \frac{1}{2}m_{sys}v_G^2 + \frac{1}{2}I_{ZZ,sys}^G\dot{\Psi}_{frame}^2 + \frac{1}{2}I_{YY,LD}\dot{\phi}_{LD}^2 + \frac{1}{2}I_{YY,RD}\dot{\phi}_{RD}^2 + + \frac{1}{2}I_{YY,LC}\dot{\phi}_{LC}^2 + \frac{1}{2}I_{YY,RC}\dot{\phi}_{RC}^2 + \frac{1}{2}I_{ZZ,LFLC}\dot{\Psi}_{LC}^2 + \frac{1}{2}I_{ZZ,RFRC}\dot{\Psi}_{RC}^2 \quad \textbf{(7)}$$

where, $\boldsymbol{m_{sys}}$: mass of the chair and occupant; $\boldsymbol{I_{ZZ,sys}^G}$: yaw moment of inertia of the wheelchair about its center of mass; $\boldsymbol{I_{YY,LD}}, \boldsymbol{I_{YY,RD}}$: moment of inertia of the left and right drive wheels about their axles; $\boldsymbol{I_{YY,LC}}, \boldsymbol{I_{YY,RC}}$: moment of inertia of the left and right caster wheels about their axles; $\boldsymbol{I_{ZZ,LFLC}}, \boldsymbol{I_{ZZ,RFRC}}$: yaw moment of inertia of the left and right caster forks about their stems. The preliminary results were demonstrated in **Table 5**. In summary, the modified figure-8 with 540° turn course includes all three types of KE especially with around 23% of the total KE in yaw (turning) energy. Therefore, for the purpose of highlighting changes in directions, we selected *modified figure-8* with 540° turn course as our wheelchair maneuver course. The use of a modified figure-8 maneuver requires that the operator change speeds and directions and overcome translational, rotational, and turning inertia as well as rolling resistance and tire scrub.

Table 5 Propulsion course comparison using TiLite Aero Z with the same configuration (front axle, spoke wheels with pneumatic tires, and 75psi tire pressure)

	Avg Vel [m/s]	**Avg tKE [J]**	**% KE in translation**	**% KE in turning**	**% KE in rotation**
Modified figure-8 with 540° turn and 2m straight distance [n=4] (M±SD)	**0.58±0.01**	**22.91±2.22**	**73.9±1.4%**	**22.9±1.5%**	**3.2±0.2%**
Modified figure-8 with 180° turn and 2m straight distance [n=1]	0.74	33.0	85.6%	11.4%	3%
Slalom course [n=6] (M±SD)	0.91±0.10	8.82±1.65	79.2±0.8%	6.4±0.4%	14.4±0.5%
1 push (2m), stop, 1 push, and turn [n=8] (M±SD)	0.33±0.07	10.90±3.18	86.4±2.8%	10.3±2.6%	3.3±0.5%
2 pushes (7m), stop, 2 pushes, and turn [n=8] (M±SD)	0.49±0.10	17.73±5.34	93.5±1.3%	3.4±1.0%	3.1±0.5%

Note: Avg. averaged; Vel: Velocity; tKE: total kinetic energy

Modified figure-8 course. The course was marked in a straight line 1.9 m apart (**Figure 15**). Subjects were instructed to follow the straightaway paths (around 2 m) to a rotation point and then performed a fixed-wheel turn of 540°. The turning radius was around 1.5m depended on the wheelchair size. Subjects continually traveled the course for five minutes to reach the steady-state O_2 consumption. A visible clock allowed subjects to maintain a consistent speed at 0.6 m/s, which reflects that of typical everyday mobility [3]. The rate perceived exertion (RPE) after the long bout maneuver was recorded immediately after the trial.

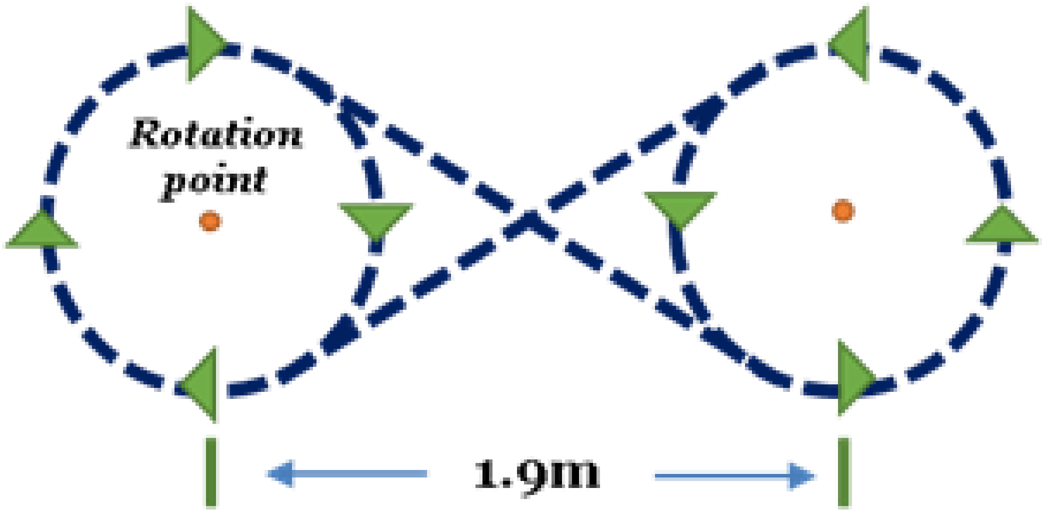


Figure 15. Modified figure-8 maneuver

Propulsion effort. To discern the cost of transportation across configurations, we used the ***propulsion effort*** to represent the amount of metabolic cost required by users over the same propulsion course. To measure propulsion effort, we asked users to execute a long bout maneuver, which is necessary because of the slow response of the cardiorespiratory system. The metabolic demand is reflected in oxygen consumption (VO_2) during and after physical activity. In the over-ground maneuver with subjects wearing a mask, we continuously measured subjects' VO_2 from expired gas using a portable VO_2 measurement system (Fitmate Pro, Cosmed) and represent their aerobic metabolic by VO_2 expressed in ml/kg/min during the task (**Figure 16**). Then we averaged the last minute of steady-state oxygen consumption to represent the metabolic demand for the over-ground maneuver (**Figure 17** and **Figure 18**).

Figure 16. Subject performing a long-bout maneuver with equipment

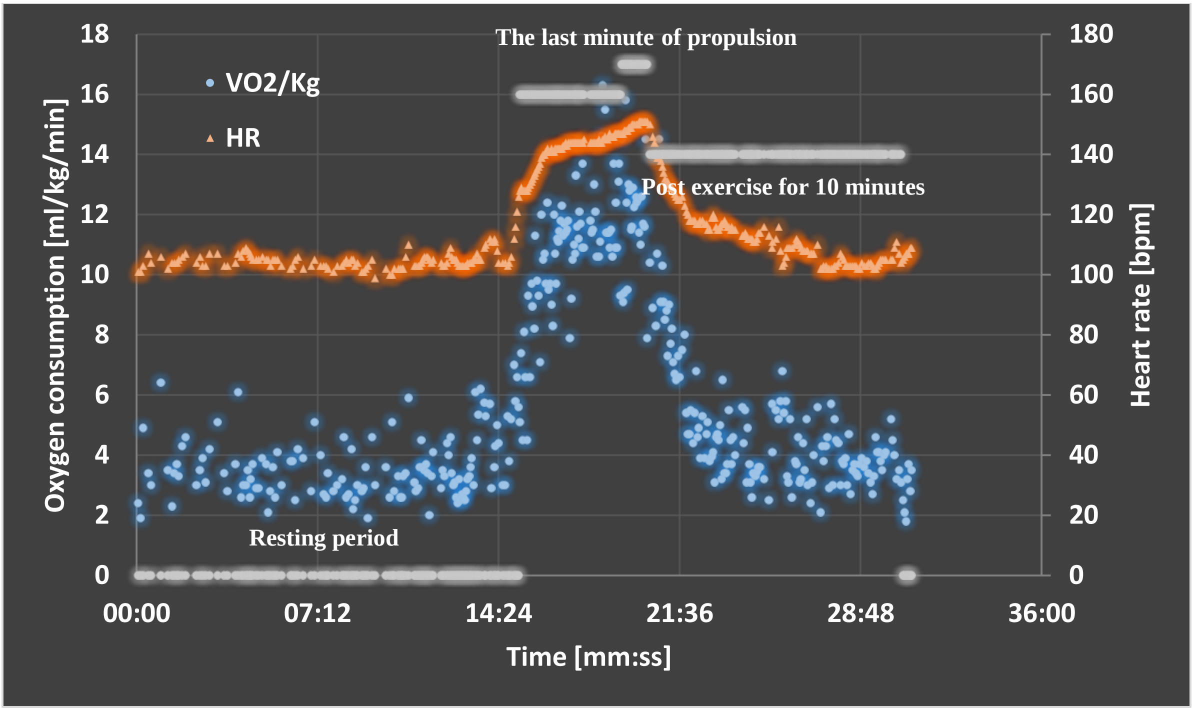


Figure 17 Physiological response during 5 minutes activity from one wheelchair user (SCI, T12)

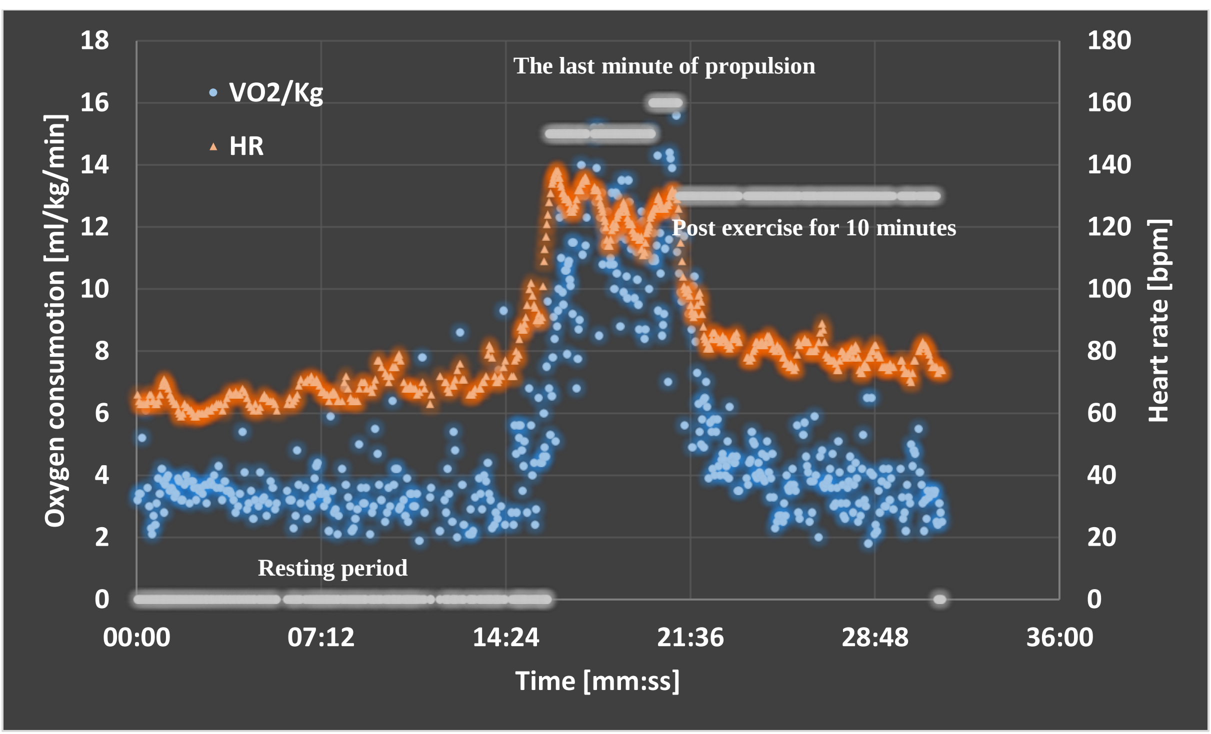


Figure 18 Physiological response during 5 minutes activity from one able-bodied subject

Subjects. Testing the propulsion effort of various wheelchair configurations requires consistent propulsive forces across test configurations. For the purpose of developing valid methodologies and evaluating the preliminary results, we recruited both able-bodied subjects and full-time MWU. The use of able-bodied subjects within wheelchair research has been identified as a useful strategy [52].

Protocol. We recruited two able-bodied subjects for testing the reliability of propulsion course. Each subject tested on two types of wheelchairs for two days. The protocol was repeated three times on the same day. Each testing was separated at least 10 minutes resting period. Kinetic energy with three components (translational, rotational, and turning energy), propulsion speed, and metabolic costs covered VO_2 and heart rate were used as the dependent measurements for testing reliability.

For testing the sensitivity in discerning propulsion efforts across wheelchair configurations, we recruited five subjects (one full-time MWU and four able-bodied subjects) to try different wheelchair configurations. The same subject tested different wheelchair configurations either on the same day with at least 10 minutes resting period or different days. Wheelchair configurations including different frame types (folding standard wheelchair, Invacare Tracer and rigid ultralightweight wheelchair, TiLite Aero Z), wheelchair masses (+0kg vs. +5.5kg) and tire inflation (100% vs. 50% tire inflation). The configurations were chosen to change the inertial and frictional parameters of the wheelchair. 5.5kg is the averaged weight difference between standard (K0001) and ultralightweight (K0005) wheelchair. ***Total oxygen consumption (**VO_2**)*** and rate perceived exertion (RPE) were used as the dependent measurements for testing sensibility.

Statistics. To assess the quality of the propulsion course, we evaluated the test-retest reliability and the test sensitivity in discerning propulsion efforts across wheelchair configurations. Descriptive statistics were used to analyze the values of propulsion effort and kinetic energy. The coefficient of variance (CV) was used to evaluate how repeatable in measuring propulsion effort and kinetic energy. Due to the small size, the study only demonstrated the difference by showing the mean value and % change.

4.1.2 Results

<u>Test-rest reliability of the modified figure-8 maneuver</u>

The results showed that the maneuver exhibited high repeatability on propulsion speed (CVs<3%), total KE imparted (CVs<3%), the last minute of VO_2 (CVs<11%), and heart rate (CVs<4%). **Table 6** is the description result of each dependent variable.

Table 6 Outcome measurement (mean±SD)

		Trial 1	**Trial 2**	**Trial 3**
Kinetic energy				
	Translational [J]	19.44±2.33	19.46±2.46	20.05±2.02
	Rotational [J]	0.84±0.10	0.85±0.10	0.87±0.08
	Turning [J]	6.83±1.48	6.91±1.42	7.02±1.48
	Total [J]	27.12±3.84	27.22±3.90	27.94±3.46
Speed	[m/s]	0.58±0.01	0.58±0.01	0.59±0.01
Total O_2 consumption	VO_2 [ml/kg/min]	10.1±1.1	9.8±1.8	9.7±1.9
	Heart rate [bpm]	85±7	84±9	82±9

Sensitivity in discerning propulsion efforts across configurations

Subjects propelling a standard folding wheelchair required 35% higher total oxygen consumption and 14% higher heart rate than they did propel a rigid ultralightweight wheelchair with the same speed, indicating a worse economy. The RPE also followed the same trend in which subjects using a standard folding wheelchair perceived 29% higher effort than they did use a rigid ultralightweight wheelchair. **Table 7** shows the description of metabolic costs with different wheelchairs.

Table 7 Metabolic effort of wheelchair propulsion

	Standard folding wheelchair	**Ultralightweight rigid wheelchair**
Total O_2 consumption [ml/kg/min]	11.3±0.6	8.4±0.4
Heart rate [bpm]	89±2	78±8
RPE	9±3	7±2

To further evaluate the sensitivity in discerning total oxygen consumption across configurations, the testing wheelchair was changed either in wheelchair masses (+0kg vs. +5.5kg) or tire inflations (100% vs. 50% tire inflation). The results demonstrated that changing inertial parameters by increasing 5.5 kg would increase 4% in total propulsion efforts, whereas changing frictional parameters by reducing 50% tire pressures would increase 12% total oxygen consumption (**Figure 19**).

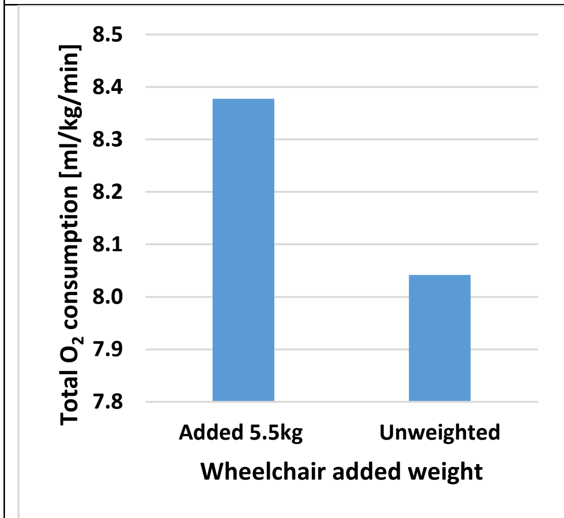
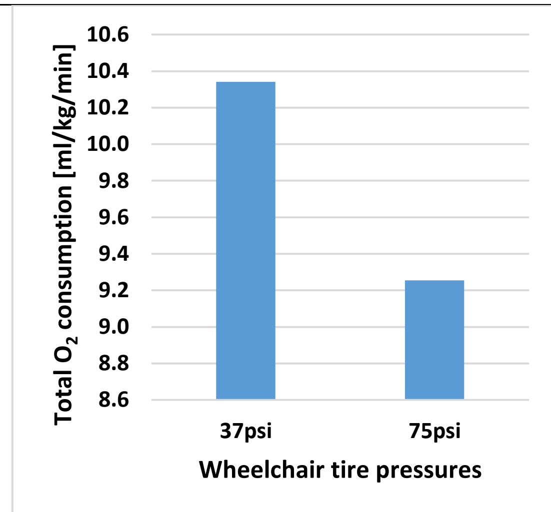
Change inertial parameter (wheelchair mass) (n=1, able-bodied)	Change frictional parameter (tire inflation) (n=1, MWU)
	

Figure 19 The cost of propulsion of MWU and able-bodied subject using different wheelchair configurations

4.1.3 Discussion

Propelling wheelchairs on the modified figure-8 course highlights the features of acceleration, deceleration, and straight/turning movements, which are reflected in daily activities. This method provides a valid mean to study metabolic efforts of wheelchair propulsions by providing a repeatable and controlled environment. In addition, the protocol can distinguish metabolic costs while changing the inertial and frictional parameters of wheelchairs.

Compared to previous kinetic energy studies of free-wheeling maneuvers [107], the modified figure-8 course had a greater percentage of turning kinetic energy (23%) than the finite radius turn (14%) and slalom course (6.4%). The modified figure-8 course had a much smaller turning ratio of KE than a zero radius turn (71%). Since propulsion speed is one of the factors that influence metabolic efforts [109, 110] and push forces [111], we controlled the propulsion speed instead of the self-selected speed to reduce confounding factors. By using the countdown clock for visual feedback, the propulsion speed was well controlled at 0.58±0.01 m/s, which falls within the daily average speed ranging from 0.5 to 0.8 m/s [3]. Propelling wheelchair in low speed can also help to reduce body sway, which might influence wheelchair maneuver due to the change of weight distribution.

Propelling on the modified figure-8 course is simple and repeatable. Previous studies using wheelchair circuit courses have shown a high intrarater reliability in measuring the sum of ability scores [103, 104], the sum of performance time [103, 104], and peak heart rates [97, 103]. Compared to previous courses, the modified figure-8 course also provides a reliable measurement of kinetic energy and metabolic costs with low coefficients of variation (CVs<11%).

By using the modified figure-8 course with controlled speeds, subjects showed different metabolic costs across wheelchair designs, including configurations of frame types, masses, and tire pressures. Compared to subjects using standard wheelchair frames, subjects using ultralightweight wheelchair frames tended to have quicker self-selected speeds [112], longer distances [112], and greater comfort [98]. However, the difference in energy costs was not clear. Paraplegic subjects using ultralightweight wheelchairs were shown to reduce oxygen costs by 11% to 17% when compared to subjects using standard wheelchairs [27, 112]. However, for tetraplegic users, Beekman et al. [112] found no difference in oxygen costs between subjects using ultralightweight and standard wheelchairs. In our study, we found that subjects needed extra oxygen cost (increased 35% on total oxygen consumption and 14% of heart rate) while propelling standard wheelchair frames compared to ultralightweight wheelchair frames on the modified figure-8 course. The trend of reducing metabolic effort while using ultralightweight wheelchair was consistent with previous wheelchair studies. In this study, the greater percentage difference across wheelchair frames on metabolic costs was probably due to different wheelchair selection or propulsion course.

Regarding adding weight to wheelchairs, Cowan et al. [22] found that users would decrease self-selected speeds but increase pushing force when adding 9.05kg to wheelchairs when propelling straight ahead. In detail, subjects reduced 4% propulsion speed but increased 5% resultant force on the tile surface, whereas reduced 3% propulsion speed but increased 4% resultant force on the low-pile carpet. However, by studying efforts on the sidewalk, stop-and-go, slalom, and treadmill, Sagawa et al. [113] found no significant effect of mass (add 0, 1, 2, 5kg) on energy expenditure. de Groot et al. [26] also found no effect of adding weight (0kg, 5kg, and 10kg extra) on power output, propulsion technique, and metabolic efforts when propelling wheelchairs on the treadmill with a fixed speed. Compared to previous studies, our preliminary results showed that the subject increased 4% on total oxygen consumption when added 5.5 kg on the wheelchair. Although we saw a small increase in oxygen consumption while adding mass on wheelchairs, we were not able to draw a conclusion due to the small sample size.

Regarding the impact of tire pressures on wheelchair propulsion, previous studies showed that subjects propelling wheelchairs with 25% tire pressure significantly increased oxygen consumption by 8% to 24% compared to 100% tire pressure in straight maneuver [26, 30]. Although de Groot et al. [26] found no significant difference in oxygen costs, Sawatzky et al. [30] found a 12% increase in oxygen costs when propelling wheelchairs with 50% than 100% tire inflation. Compared to Sawatzky's et al., [30] study, our preliminary results showed a similar result that the subject needed extra 12% total oxygen consumption while propelling the wheelchair with 50% tire inflation compared to the subject propelling with 100% tire inflation. The trend of saving metabolic efforts while using a wheelchair with inflated tires was consistent with previous wheelchair studies.

Subjects need to accelerate, decelerate, and turn while maneuvering on the modified figure-8 course. Propelling wheelchairs on the selected course is simple and repeatable. By controlling the propulsion speed, we were able to find a difference in metabolic costs when subjects used different wheelchair frames, wheelchair masses, and tire pressures. Since propelling wheelchairs on the modified figure-8 course is reliable and valid, we used this propulsion course as the standard approach to evaluating the performance of wheelchair propulsion for our subsequent study.

4.2 Aim #3. 2: To characterize wheelchair users' physical and biomechanical variables and determine their relationship to propulsion efforts

4.2 Aim #3. 3: To identify the mechanical parameters of the wheelchair and determine their relationship to propulsion efforts

4.2 Aim #3. 4: To evaluate the combined impact of operator and mechanical factors on propulsion efforts

Wheelchair users vary widely in their level of cardiorespiratory and musculoskeletal fitness, some being seriously unfit and others having levels that compare closely with those of fit able-bodied athletes [40]. It is generally accepted that active MWU are healthier than non-active MWU. However, the rates of physical inactivity among wheelchair users remain extremely high [114]. Only 27% of adults with physical disabilities engage in regular, moderate aerobic physical activity and a mere 15% participate in muscle-strengthening activates at least two days each week [115]. The consequences of inactivity in wheelchair users are profound and may lead to a host of complications, including muscle atrophy as well as increased risk of overuse injuries and upper limb pain. The occurrence of these conditions may lead to further disabilities by contributing to decreasing in mobility and physical function [115]. Building muscle strength and aerobic capacity are both important for wheelchair users for improving health conditions. These two physical characteristics are even correlated with each other [41, 84]. Referring to Kofsky's et al. [84] study, they found that arm strength has more impact upon cardiovascular performance than the anatomical level of the lesion. However, the extent to which fitness factors have a greater influence on the propulsion performance is less clear. In other words, it is unclear that whether cardiorespiratory fitness like aerobic capacity or peripheral factors like muscle strength would improve propulsion economy. Although the distance between shoulder and wheelchair axle has been shown to influence wheelchair propulsions, it is still unknown what is the related influence of biomechanical and physical factors on propulsions. Therefore, the goal for Ai 3.2 is to build a regression model that included both biomechanical and physical factors in predicting propulsion efforts.

Although many studies have shown that wheelchair designs could influence wheelchair propulsion in a different manner, limited studies have investigated the related influence of mechanical properties on wheelchair propulsion, especially for daily maneuvers. In addition, no studies have investigated the combined effect of operator and mechanical factors on wheelchair propulsion. Therefore, the overall goal for Aim 3.3 is to build a regression model that identifies the impact of mechanical factors on propulsion efforts. In addition, Aim 3.4 summarizes the results from Aim 3.2 and Aim 3.3 and builds a regression model that considers both the influences of operator and mechanical factors on propulsion efforts.

4.2.1 Methods

<u>Subjects</u>

Able-bodied subjects. This study of wheelchair propulsion seeks to investigate wheelchair mechanical parameters as well as operator biomechanical and physical variables. Its design, therefore, aims to incorporate a range of both sources of variance. Able-bodied subjects, considered novice (or untrained) in wheelchair propulsion, will have a homogeneous characteristic of full function in their upper extremities and trunk musculature. Therefore, we recruited healthy adults (18-60 years old, both male and female) with no upper extremity limitation or injury.

Manual wheelchair users. This study included all MWU instead of users with specific diagnoses. People who are independent in propulsion are typically prescribed K0004 and K0005 classes of wheelchairs. According to Medicare wheelchair purchases, the top five diagnostic categories for these chairs are strokes, neurologic disorders, osteoarthritis, heart disease, and spinal cord injury. In addition, classifying wheelchair users based on their functional ability (e.g., ISMGF) is standard for athletic events. Therefore, limiting our study group to a particular diagnosis would exclude MWU who might benefit from the study objective. The study recruited adult (18-60 years old, both male and female) MWU who have used wheelchairs as their primary means of mobility for at least six months, based on the assumption that, after six months, they have settled into a consistent pattern of propelling and maneuvering their wheelchairs. Because of the requirements of predicting peak VO_2, we only included users who could perform arm crank ergometry at least 12 minutes without interruption at a loading similar to that of over-ground manual wheelchair propulsion. The study excluded people 1) with an upper extremity limitation or injury that precludes bilateral propulsion of a manual wheelchair, 2) with a history of cardio-respiratory disease that may impact extended bouts of propulsion, and 3) who are unable to provide consents.

Protocol

The protocol was approved by Georgia Tech Institutional Review Board (IRB). Subjects made two visits to complete the experiment. Subjects were asked not to eat or drink any caffeinated drinks 3 hours before the testing. During the first visit, subjects completed coast-down tests [116] and long-bout maneuvers on both tile and carpet surfaces. During the second visit, subjects completed the maximum isometric test and multistage submaximal arm exercise. Each visit was completed within 2.5 hours. In this experiment, MWU used their wheelchairs while able-bodied subjects were assigned to randomly pre-configured wheelchairs, which varied in tire types, axle positions, and frame types.

We measured inertial and frictional parameters by a dynamic analysis and a coast-down protocol (Aim # 1.1), propulsion strength by the modified isometric propulsion test as subjects sit in their wheelchairs (Aim # 2.1), aerobic capacity by a multistage incremental exercise with the arm ergometry, and shoulder position by a digital photographic technique. We recorded propulsion efforts in real-time during the modified figure-8 maneuver (Aim # 3.1).

Instrumentation

Metabolic system. The propulsion demand is reflected in oxygen consumption (VO_2) during over-ground maneuvers. We continuously measured subjects' VO_2 from expired gas using a portable VO_2 measurement system (Fitmate Pro, Cosmed, Italy) and represent their aerobic metabolism by VO_2 expressed in ml/kg/min during the task. Breath-by-breath data were averaged over the last minute of VO_2 to represent the propulsion demand for the over-ground maneuver. We also used the metabolic system for evaluating aerobic fitness by doing a multistage arm-cycling exercise protocol. The flow turbine was calibrated with a 3-L syringe before every test. The O_2 sensor was calibrated automatically during the test (the interval time is 6 minutes by default).

iMachine. System mass (kg), rotational moment of inertia (Izz), and weight distribution (%) were measured experimentally using a device called the iMachine [67]. The device consists of a turntable mounted to a single axle. Load cells mounted on the turntable measure the mass and CoM of the wheelchair. An encoder measures the rotation of the turntable, whose oscillations are damped by a spring of known stiffness. The detail in calculating weight distribution was shown in the Data Analysis section.

Accelerometers. To measure overall friction, each participant's wheelchair was outfitted with a data logging system on both drive wheels for measuring the wheel rotation rate [3]. The data logging system (MSR 145, Swiss) features a solid-state, triaxle accelerometer with a ±1 g range at its core with a 50 Hz sampling rate.

Surface condition. Each subject propelled his/her wheelchair on an indoor surface comprised of linoleum tiles and low-pile carpet. In addition, all coast-down tests were conducted on the same surfaces. The standardized coefficient of kinetic friction of the tile and carpet surface were found to be 0.92 and 1.11, respectively using RESNA's test procedure to characterize surfaces during wheelchair testing [117].

Over-ground maneuver. Subjects were instructed to follow the straightaway paths (around 2 m) to a rotation point and then perform a fixed-wheel turn for 540° (Figure 15). The use of a modified figure-8 maneuver requires that the operator change speeds and directions and overcome translational, rotational, and turning inertia as well as rolling resistance and tire scrub [107]. Subjects continually traveled the course for five minutes. A visible clock allowed subjects to maintain a consistent average speed at 0.6

m/s. The target speed was chosen to reflect typical everyday mobility [3]. The detail of course design was described in Aim # 3.1.

Data Analysis

Outcome measurements

Net propulsion effort. It is a construct that best reflects the required metabolic effort of performing a “task.” In other words, the measurement quantifies the metabolic cost of propelling manual wheelchairs to complete such maneuver. The greater the effort of propulsion, the less economy a wheelchair is, for it requires more energy/effort to perform the same task. According to Hintzy et al. [118] study on mechanical efficiency, the analysis method using resting metabolism as the baseline correlation will be able to include all unmeasured work components (e.g., skeletal muscle activity and organ system adjustments during unloaded movement of the limbs, unaccounted muscles involved in body stabilization, transfer of force from skeletal muscles to the propulsion system, and isometric exercise) [119]) except for resting metabolism. Using resting metabolic rate as a correlation factor was also suggested to adjust for individual differences when evaluating the energy cost of walking [120]. Therefore, to capture the actual value of metabolic costs for wheelchair propulsions, we used net propulsion effort as our dependent measurement. Metabolic energy costs (VO_2) of the wheelchair maneuver will be measured in ml/kg/min normalized body mass. The resting metabolic rate was measured by averaging the last two minutes of VO_2 during at least 10 minutes’ rest.

Operator factors: Physical fitness factors

Aerobic capacity. Maximal oxygen uptake (peak VO_2) is the maximum rate of oxygen consumption, which typically reflects the aerobic capacity (physical fitness) of the individual. However, the traditional protocol of measuring peak VO_2 is time-consuming and required a high degree of compliance from the individual involved. Subjects with injuries or paralysis of their lower limbs cannot have their aerobic fitness evaluated through cycling or running bur arm cranking. There are also practical concerns when doing exhaustive exercise with non-athletic or patient population [121]. Even with modified field protocols [122, 123], the maximal energy expenditure may be influenced by wheelchair related factors including rolling resistance, wheelchair configurations, and propulsion skills [124]. According to the reason of safety, reduced fatigue, and independent to wheelchair related factors, we decided to use sub-maximal arm exercise procedure to predict peak VO_2 for MWUs.

Using the rating of perceived exertion (RPE) during a sub-maximal exercise is widely used to predict peak VO_2 [125-128], even for people with SCI [125, 127]. According to Goosey-Tolfrey's et al.[125] study in trained wheelchair sports persons, they have used differentiated RPE, which includes central, peripheral, and overall RPE, from each exercise stage to predict peak VO_2. The results showed that no significant difference was found between measured and predicted peak VO_2. Using RPE measurements have an even better prediction for peak VO_2 than using heart rate [125].

To evaluate aerobic capacity, we asked subjects to perform arm ergometer using three stages of fixed power output. Resting metabolic rate was measured for 5 minutes once the subjects have been sitting quietly for 10 minutes. During the arm exercise, a participant was seated so the axis of arm ergometry is positioned level with the shoulder joint, positioned a distance from the arm ergometry, which allowed for a full-arm extension during the crank rotation [129]. The multistage exercise activity commenced with 10W power output with 10W increments to 30W (maximum) at 70 rpm. Each stage continued for 3 minutes followed by at least 1 minute of resting before progressing to the next stage. We chose this range because it reflects effort expended during typical wheelchair propulsion [130]. According to a review paper on energy costs during a variety of physical activities of MWU, Conger et al. [130] reported that the effort expended in 16W arm ergometry was equivalent to during propulsion on tile and ergometry at 32 W was consistent with propelling on the sidewalk. According to previous pilot data, this range of exercise covered the exercise intensity of MWU propelling during the modified figure-8 maneuver for five minutes. Metabolic responses (VO_2 and HR) were continually measured during the whole exercise. RPE was reported during the resting period. The detailed protocol is described in **Appendix A**. Three data points of VO_2 in reflected RPE from three stages arm exercise were used to build a linear regression line. Using individual participant linear regression, VO_2 was regressed against the corresponding RPE values and extrapolated to the theoretical maximal RPE (RPE 20) on Borg scale to predict peak VO_2 (ml/kg/min). The predicted value was used to reflect the physical fitness of subjects. **Figure 20** is the example of predicted aerobic capacity from two MWU and one able-bodied subject.

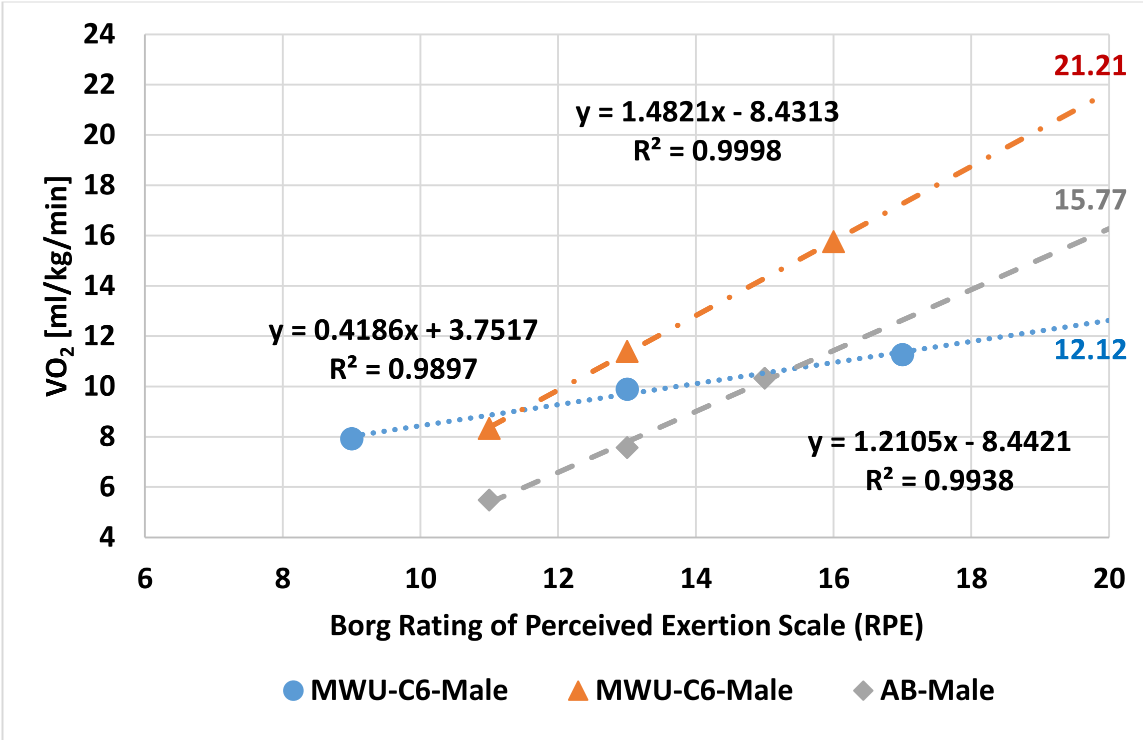


Figure 20 Metabolic response in graded-arm exercise

Propulsion strength. The detail of strength measurement is described in Aim #2.1. To measure the maximum propulsion strength, all wheelchair users sit on their wheelchairs, whereas able-bodied subjects sit on wheelchairs randomly selected. After attaching the wheelchair on the designed platform, all subjects grabbed the top dead center of the wheelchair push-rim and then push the push-rim forward as hard as they can. The pushing consisted of 5s maximal isometric force exertions of both arms together with four repeated trials. To minimize the influence of body mass as an underlying reason for the relations studied, all strength measurements were normalized and expressed as force per unit body mass (N/kg).

Operator factors: Biomechanical parameters

Shoulder position. The related distance between the shoulder and axle positions was used to represent the shoulder/axle position as a biomechanical measurement. To measure the related distance, we will deploy the digital photographic technique, which has been widely used in human body alignment and length studies [131-135]. Referring to the Niekerk et al. [135] and Boninger et al. [24] studies, we will position a digital camera mounted on a tripod and place it 2m away from the wheelchair. Subjects were instructed to sit at rest and in their normal seated positions. Hands were asked to place comfortably on the top of thighs (**Figure 21**). This position was chosen so that it could easily be copied in a clinical environment. To improve accuracy, we placed two color markers on the shoulder (acromion process) and the rear axle (hub) as references. The horizontal distance (***XPOS***) was calculated by ImageJ (Version 1.51). The farther back the axle is with respect to the shoulder, the more positive XPOS. If the axle is in front of the shoulder, XPOS is negative.

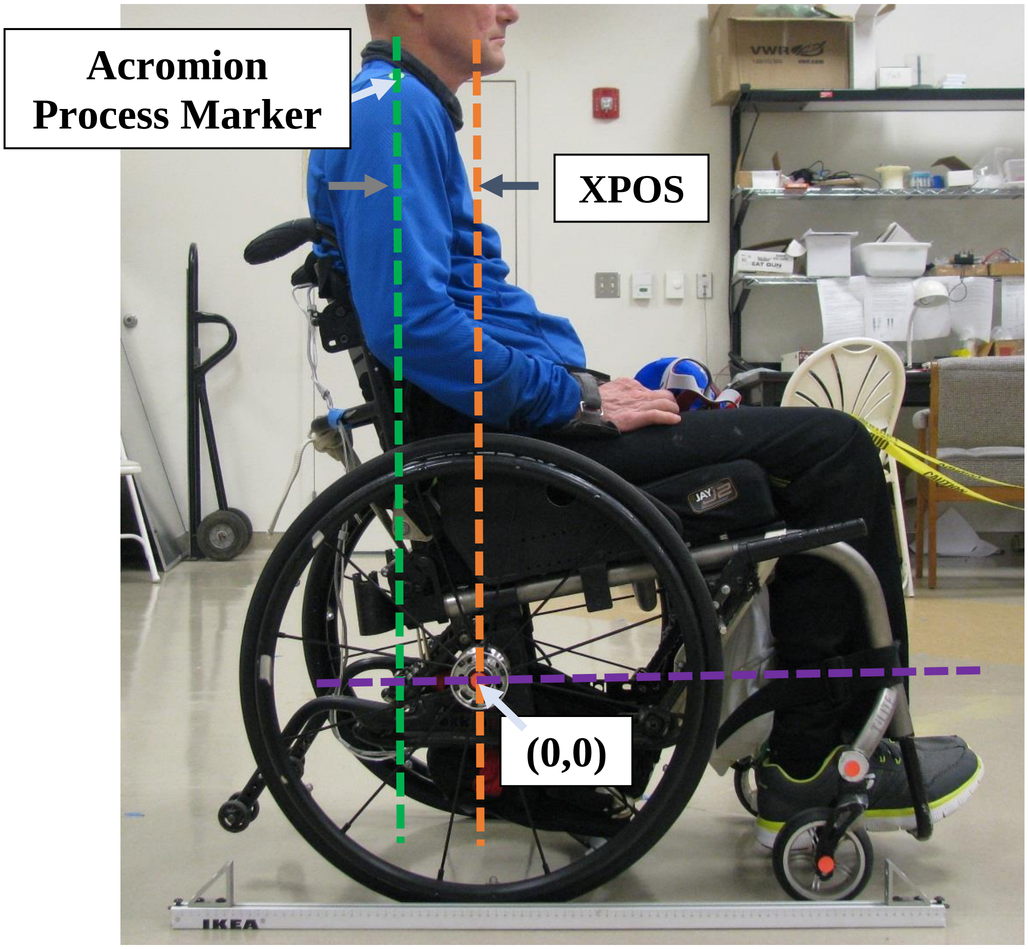


Figure 21 Example of shoulder position parameter

Mechanical parameters

System mass. The total mass of the whole wheelchair system, which included wheelchair weight, equipment weight, and subjects' body weights was measured by ***iMachine*** (**Figure 22**).

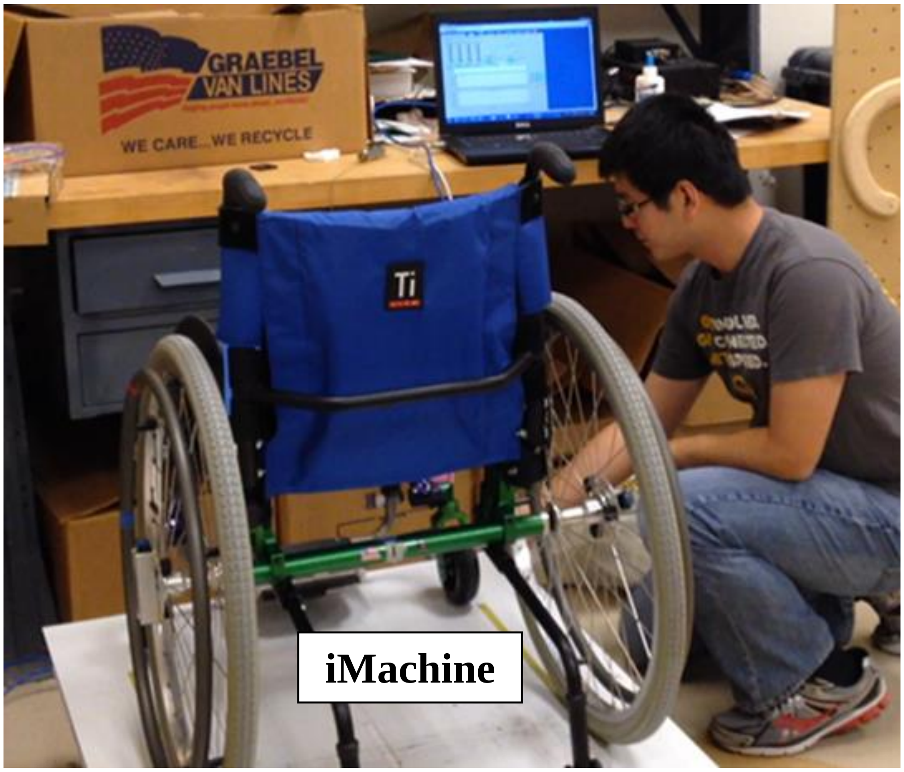


Figure 22 iMachine

Weight distribution. The weight distribution of the wheelchair system (included wheelchair and the human body) was represented as the percent (%) loading on the drive wheels. To obtain this value, we used ***iMachine*** (**Figure 22**) to measure the position of the center of mass (CoM). After aligning the CoM to the axis of rotation (**Figure 23**), we measured the horizontal distance [A] between the center of the rear axle and the axis of rotation. At the same time, we measured the horizontal distance [B] between the center of the real axle and the center of casters. According to the *law of class I lever*, the ratio of weight loading on drive wheels and casters is given by the ratio of the distances from the fulcrum to the points of contact points of wheels.

$$\frac{Md}{Mc} = \frac{(B-A)}{A}$$

$$\frac{Md}{Msys - Mc} = \frac{(B-A)}{A}$$

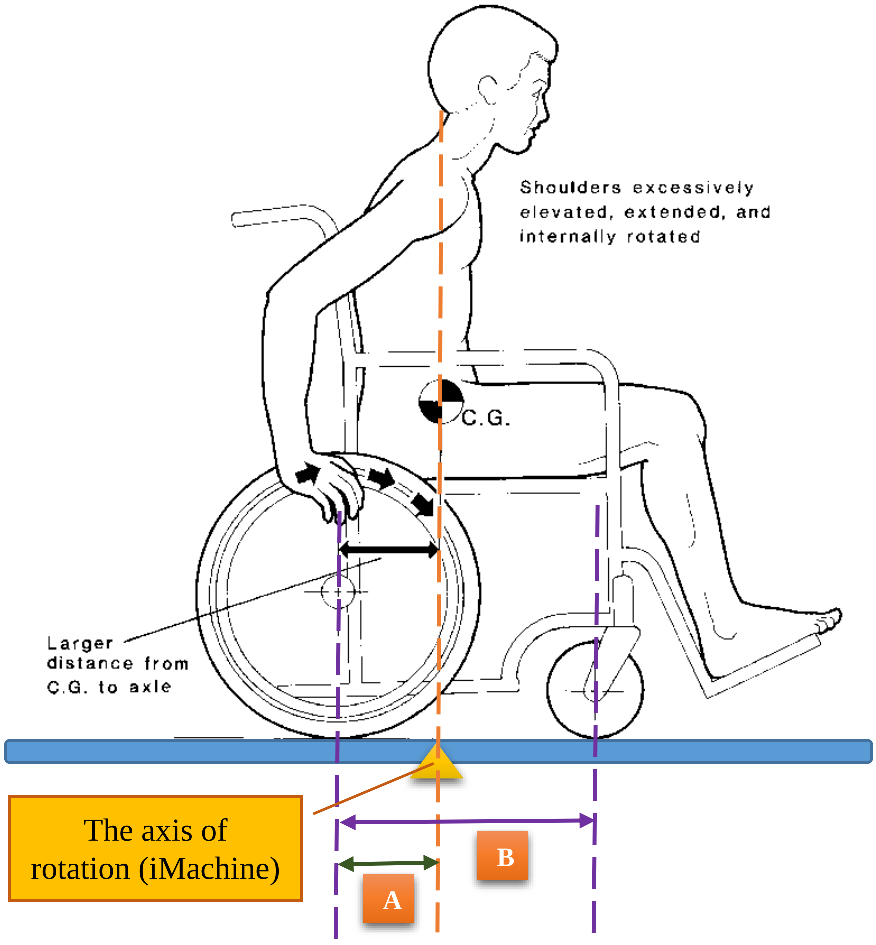
$$Md = \frac{(B-A)}{A} \times (Msys - Mc)$$

$$(1 + \frac{(B-A)}{A}) \times Md = \frac{(B-A)}{A} \times Msys$$

$$\frac{Md}{Msys} = (\frac{B}{A} - 1) \times \frac{A}{B}$$

$$\%\ loading\ on\ drive\ wheels = \left(1 - \frac{A}{B}\right) \times 100\%$$

where M_d is the loading on drive wheels (kg), M_c is the loading on casters (kg), M_{sys} is the system mass (kg).



Note: A: horizontal distance from drive wheels to casters (m); B: the distance of center of mass location forward from the center of rear axle (m)

Figure 23 the weight distribution of the wheelchair system

Frictional parameters. A coast-down protocol was deployed to measure frictional energy loss during straight trajectories and fixed-wheel turns [136]. Decelerations (m/s^2) recorded through accelerometers [3] on both wheels were averaged to represent frictional parameters from each direction. Measurements from left and right turns were averaged to provide a single turning value. Post processing of all deceleration values was fed into custom-made software (Matlab 2013, MathWorks, USA). The detail of coast-down protocol was described in Aim #1.1. Since rolling resistance and scrub torque are function of mass and inertia separately, this study used deceleration values to represent frictional parameters.

Statistics

Descriptive statistics were calculated for the demographic and clinical characteristics of subjects. To assess the normality of data below 50 samples [137], this study applied Shapiro-Wilk Test and Normal Q-Q Plot to check the assumption of parametric testing. Since normality was rejected ($P < 0.05$) for the measurements of muscle strength, aerobic capacity, and turning deceleration of our samples, a non-parametric statistical analysis was used in this study.

Nonparametric Spearman rank-order correlations measured the strength and direction of the proportional relationship between propulsion efforts and predictors. Each bivariate scatterplot diagram was generated to check potential outliers. The absolute correlation coefficient values (γ) were interpreted according to the guidelines proposed by Altman [66]: poor (γ<0.20), fair (0.21-0.40), moderate (0.41-0.60), good (0.61-0.80), or very good relationship (0.81-1.00). The eligible modifiable determinants based on the research questions were entered into a separate linear multiple regression analysis with backward selection techniques. The possible prediction equations were developed based on the best two predictors by backward selection techniques. A separated adjusted R^2 value was reported for each net propulsion effort (ml/kg/min) as a conservative estimate of the strength of the regression given the number of predictors (operator and mechanical factors) considered by each model. All statistical analyses were computed with SPSS version 22.0 for Windows (SPSS, Inc., Chicago, IL). The probability level of p equal to

0.05 or less was accepted as statistically significant. The following are predictor variables used specifically for Aim #3.2 and Aim #3.3. For Aim #3.4, we chose the most significant predictor from the operator and mechanical prediction models, respectively, to build a combined regression model.

Aim 3.2 (biomechanical and physical parameters)

1) **XPOS (cm)**: the horizontal position of the axle with respect to the shoulder girdle
2) **Isometric strength (F_{iso}, N/kg)**: the averaged maximum upper extremity strength across the left and right sides
3) **Aerobic capacity (ml/kg/min)**: predicted by the submaximal incremental arm exercise

Aim 3.3 (mechanical parameters)

1) **System Mass (M_{sys}, Kg)**: the total mass of the wheelchair system
2) **Weight distribution (%DW, %)**: the percentage of system mass on drive wheels
3) **Deceleration-straight (m/s^2)**: the frictional energy loss using the straight coast-down test on the tile and carpet
4) **Deceleration-turning (m/s^2)**: the frictional energy loss using the turning coast-down test on the tile and carpet

Aim 3.4 (combined parameters)

1) **Aerobic capacity (ml/kg/min)**: predicted by the submaximal incremental arm exercise
2) **Weight distribution (%)**: the percentage of system mass on drive wheels

4.2.2 Results

Subjects

Thirty-six individuals, recruited through fliers, direct contact, and previous study participation, volunteered for participation in this study. Within these thirty-six individuals, thirteen individuals were able-bodied and twenty-three were full-time wheelchair users. In wheelchair user group, one subject with symptom ataxia, and twenty-two subjects were SCI. All thirteen able-bodied subjects completed the test, but one wheelchair user wasn't able to finish the propulsion course in the required time. After 5 minutes propulsion in moderate speed, subjects' averaged RPE on tile surface (11.4 ± 2.0) was significantly lower than subjects reported on the carpet surface (13.4 ± 2.0), $p<.001$. Participants' outcome measurements are presented in **Table 8**. Although the subjects' injury type and level would influence their functional outcome [43], this study did not find a significant difference in physiological and biomechanics measurements (**Table 9**). Therefore, to increase statistical power with bigger sample size and have a generalized result, this study combined both able-bodied and MWU group for later data analysis.

Table 8 Subject characteristics

Characteristics	**Able-bodied [N=13]** 5 females, 8 males	**MWUs [N=23]** 2 females, 21 males
Age (yr)	24±7	40±10
Weight (kg)	75.9±17.8	74.3±20.0
Height (cm)	171.4±8.4	180.0±9.8
Time Since Injury (yr)	N/A	11.0±8.3
Type & Level of Injury	N/A	Ataxia: N=1 SCI: (C5-L2) • C-level: N=5 • T-level: N=16 • L-level: N=1

Note: Data are reported as mean ± standard deviation

Table 9 No significant difference on physiological measurements between able-bodied and manual wheelchair user group

	Able-bodied [N=13]	**MWU [N=23]**	**Sig. (Mann-Whitney test)**	**95% Confidence Interval difference**	
	Mean±S.E.	Mean±S.E.		Lower	Upper
Propulsion effort_tile [ml/kg/min]	7.4±.6	7.2±.4	.649	AB: 6.3 MWU: 6.4	AB: 8.6 MWU: 8.0
Propulsion effort_carpet [ml/kg/min]	10.6±.6	9.5±.5	.229	AB: 9.4 MWU: 8.5	AB: 11.8 MWU: 10.6
Body mass [kg]	74.7±5.4	101.9±4.3	.100	AB: 62.9 MWU: 75.5	AB: 86.5 MWU: 93.2
Shoulder position [cm]	-2.1±1.5	-2.3±1.4	1.00	AB: -5.5 MWU: -5.2	AB: 1.3 MWU: .6
Peak VO_2 [ml/kg/min]	21.2±1.2	20.3±1.9	.267	AB: 18.5 MWU: 16.3	AB: 23.9 MWU: 24.3
Strength [N/kg]	2.6±.2	3.2±.3	.159	AB: 2.2 MWU: 2.7	AB: 3.1 MWU: 3.7

Aim 3.2 (biomechanical and physical factors)

Table 10 is the detail correlation results between each operator factor. There was a strong, positive correlation between propulsion efforts on tile and carpet (γ_s = .656, p < .01). In addition, subjects' shoulder positions were positively correlated with propulsion effort on tile and carpet, γ_s = .440, p < .01 and γ_s = .427, p = .01, respectively. However, neither muscle strength nor aerobic capacity had a strong correlation with net propulsion effort.

Table 10 Spearman correlation

	Net effort Carpet	XPOS	F_{iso}	Peak VO_2
Net effort_Tile	**.656* (p=.000)**	**.440* (p=0.008)**	.023 (p=.895)	-.176 (p=.311)
Net effort_Carpet	1	**.427* (p=.010)**	.075 (p=.667)	-.054 (p=.760)
XPOS		1	.211 (p=.216)	-.064 (p=.711)
F_{iso}			1	.280 (p=.098)
Peak VO_2				1

**p*<.05

To test the hypothesis that a propulsion effort is a function of three variables, the shoulder position (XPOS), propulsion strength (F_{iso}), and aerobic capacity (Peak VO_2), a backward multiple regression analysis was performed. The results of the regression indicated that two predictors explained around 20 % of the variance (Tile: Adjusted R^2 =.197, F (2,32)=5.173, *p*=.01; Carpet: Adjusted R^2 =.194, F (2,32)=5.095, *p*=.012) (**Table 11**). The sign of the regression weights is in the predicted direction, with propulsion effort being positively associated with shoulder position, but negatively associated with aerobic capacity. However, only the shoulder position added statistically significantly to the prediction model (**Table 12**).

Table 11 Regression Model Summary

		Cumulative Adjusted R^2	Sig. for cumulative model
Net effort_Tile			
	XPOS*	.169	.008
	Peak VO_2*	.197	.011
	F_{iso}*	.172	.031
Net effort_Carpet			
	XPOS*	.217	.003
	Peak VO_2*	.194	.012
	F_{iso}*	.171	.032

**p*<.05

Table 12 Coefficient Summary

Net effort_Tile						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	8.646	.811		<.001	6.994	10.299
XPOS*	.115	.047	.386	.02	.019	.212
Peak VO_2	-.054	.037	-.231	.154	-.130	.021
						Adjusted R^2 =.197
Net effort_Carpet						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	10.546	1.024		<.001	8.461	12.632
XPOS*	.180	.060	.479	.005	.059	.302
Peak VO_2	-.013	.047	.045	.777	-.109	.082
						Adjusted R^2 =.194

**p*<.05; C.I.: confidence interval

<u>Aim 3.3 (mechanical parameters)</u>

Table 13 is the detail correlation results between each mechanical factor. Within all the mechanical parameters, only the weight distribution and propulsion efforts on either tile or carpet surface were significantly correlated. Regarding the frictional loss, there was a negative correlation between weight distribution and deceleration in turns on the tile as well as a positive correlation between weight distribution and deceleration in straight on the carpet (**Figure 24**). There was also a negative correlation between system mass and deceleration in straight on the carpet. The decelerations in straight and turning were highly correlated.

Table 13 Spearman correlation

	%DW	Decel_St	Decel_turn	Mass_sys	Inertia
Net effort_Tile	**-.651* (p=.000)**	-.214 (P=.216)	.110 (p=.529)	.179 (p=.304)	.201 (p=..248)
%DW	1	.094 (p=.586)	**-.534* (p=.001)**	-138 (p=.423)	-.205 (p=.231)
Decel_St		1	**.333* (p=.047)**	.242 (p=.156)	-.248 (p=.145)
Decel_turn			1	-.242 (p=.156)	-.100 (p=.560)
Mass_sys				1	**.905* (p=.000)**
	%DW	Decel_St	Decel_turn	Mass_sys	Inertia
Net effort_Carpet	**-.431* (p=.010)**	**-.345 (P=.042)**	-.082 (p=.639)	.024 (p=.890)	.054 (p=..757)
%DW	1	**.374* (p=.025)**	-.266 (p=.117)	-138 (p=.423)	-.205 (p=.231)
Decel_St		1	**.386* (p=.020)**	**-.368* (p=.027)**	-.297 (p=.078)
Decel_turn			1	-.003 (p=.988)	.157 (p=.362)

*p<.05

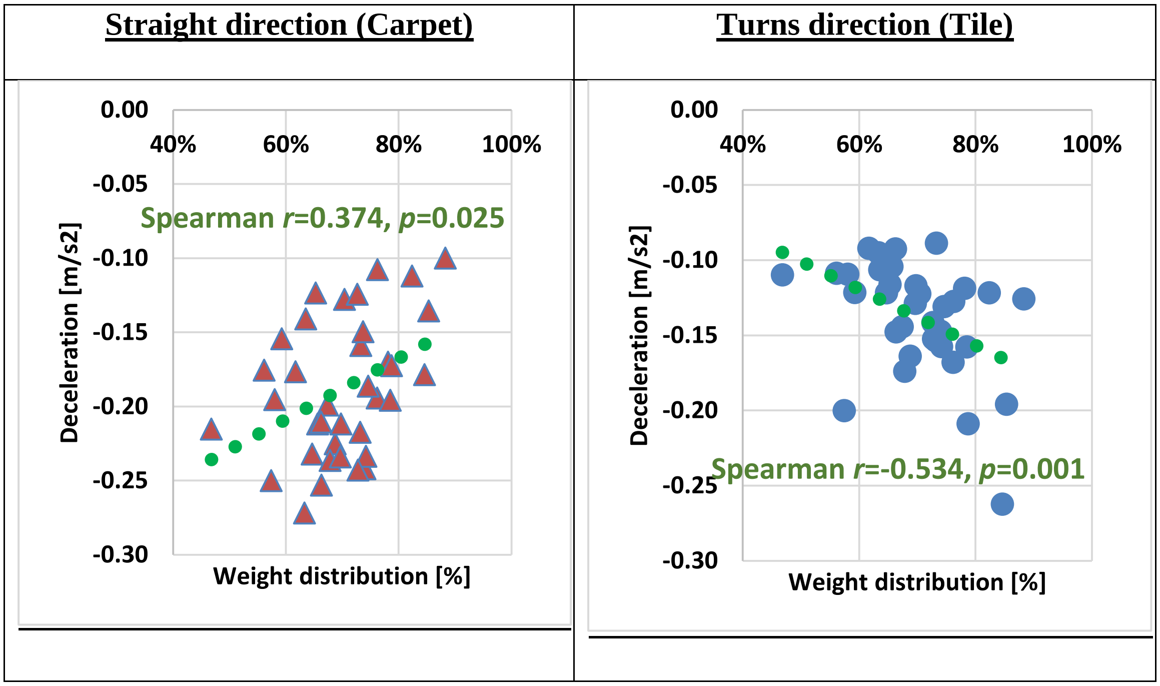


Figure 24 Correlation between deceleration and weight distribution

A two-way ANOVA was conducted for examining the effect of caster and drive wheel width on the frictional loss for each direction (straight and turn) and surface (tile and carpet). To simplify the comparison, tire width was coded with 0 and 1 (0: drive wheel or caster width < 27mm). Since friction is the function of system mass, we considered system mass as covariate. In addition, we only included ultralightweight wheelchairs (N=25) since we also want to exclude the confounding variance of wheelchair frames. When no interaction effect was found, main efforts were reported to evaluate differences between wheel widths. There was no statistically significant interaction between the caster and drive wheel width on the frictional loss. We only found a significant effect for drive wheel width during turns on the carpet $F(1, 20) = 6.947$, $p = .016$, such that friction was significantly greater for wider drive wheels (M = -.308, SE =.007) than narrower drive wheels (M = -.277, SE = .009).

Multiple regression analysis was further used to test if the weight distribution (%DW) and friction (Decel_st or Decel_turn) significantly predicted propulsion effort. The results of the regression indicated that two predictors explained 17-37 % of the variance (Tile: Adjusted R^2 =.373, F (2,32)=11.096, *p*<.01; Carpet: Adjusted R^2 =.173, F (2,32)=4.546, *p*=.018) (**Table 14**). Both weight distribution and decelerations in either straight or turning trajectory, were negatively associated with propulsion effort. However, only the weight distribution added statistically significantly to the prediction (**Table 15**).

Table 14 Regression Model Summary

		Cumulative Adjusted R^2	Sig. for cumulative model
Net effort_Tile			
	%DW*	.364	<.001
	Decel_turn*	.373	<.001
	Inertia*	.361	.001
	Mass_sys*	.370	.001
	Decel_turn	.358	.003
Net effort_Carpet			
	%DW*	.164	.009
	Decel_St*	.173	.018
	Mass_sys*	.175	.030
	Inertia	.151	.062
	Decel_turn	.129	.108

*p<.05

Table 15 Coefficient summary

Net effort_Tile						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	16.077	1.976		<.001	12.052	20.101
%DW*	-14.417	3.119	-.699	<.001	-20.771	-8.064
Decel_turn	-9.426	7.762	-.184	-.233	-25.236	6.384
	Adjusted R^2 =.373					
Net effort_Carpet						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	14.615	3.975		.001	6.519	22.711
%DW*	-9.309	4.402	-.358	.042	-18.275	-.343
Decel_St	-9.586	8.305	-.196	.257	-26.502	7.330
	Adjusted R^2 =.173					

*p<.05; C.I.: confidence interval

<u>Aim 3.4 (combined parameters)</u>

The operator and mechanical parameters, which had the highest correlation coefficient were considered for the combined regression model. According to the results from Aim #3.2 and Aim #3.3, both shoulder position and weight distribution had a high correlation with propulsion effort. However, these two variables were also highly correlated with each other (γ_s = .641, p < .01). By comparing the strength of correlation coefficient, weight distribution (**Table 13**) had a stronger correlation with propulsion efforts than shoulder position (**Table 10**). Therefore, the study selected only weight distribution for the combined regression model. Although aerobic capacity was not significantly correlated with the propulsion effort, it is the second predictor in the operator factors model (Aim #3.2). As a result, this study chose aerobic capacity as a physical predictor.

A backward linear regression was completed with the propulsion effort as the dependent variable and the mechanical (weight distribution) and physical (aerobic capacity) factors as the independent variables. The results of the regression indicated that two predictors explained 37 % of the variance on tile condition and 14% of the variance on carpet condition (Tile: Adjusted R^2 =.369, F (2,32)=10.946, *p*<.01; Carpet: Adjusted R^2 =.140, F (2,32)=3.759, *p*=.034) (**Table 16**). Both weight distribution and aerobic capacity were negatively associated with propulsion efforts. However, the weight distribution is the only significant contributor to predicting propulsion effort (**Table 17**).

Table 16 Regression Model Summary

		Cumulative Adjusted R^2	Sig. for cumulative model
Net effort_Tile			
	%DW*	.364	<.001
	Peak VO_2*	.369	<.001
Net effort_Carpet			
	%DW*	.164	.009
	Peak VO_2*	.140	.034

*p<.05

Table 17 Coefficient summary

Net effort_Tile						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	16.341	1.983		<.001	12.302	20.380
%DW*	-11.821	2.927	-.573	<.001	-17.783	-5.860
Peak VO_2	-.038	.033	-.161	.265	-.106	.030
						Adjusted R^2 =.369
Net effort_Carpet						
	Unstandardized		Standardized		95% C.I. interval for B	
	B	SE	Beta	Sig.	Lower bound	Upper bound
(constant)	17.852	2.917		<.001	11.910	23.793
%DW*	-11.006	4.305	-.424	.016	-19.776	-2.236
Peak VO_2	-.012	.049	-.039	.815	-.112	-.089
						Adjusted R^2 =.140

*p<.05

4.2.3 Discussion

Aim 3.2 (biomechanical and physical factors)

Biomechanical studies in wheelchair propulsion have been widely studied to optimize maneuver performance or prevent injury [57]. The factor of most practical significance for wheelchair propulsion was user position relative to the drive wheels [22, 24, 36]. Besides biomechanics factors, physical capacity, which includes aerobic capacity and muscle strength [40, 43, 138], were also used for the evaluation of fitness or therapeutic interventions [56].

According to our results, only shoulder positions were significantly and positively correlated with propulsion efforts on either tile or carpet surfaces. The regression model of the top two predictors for operator factors further supported this finding. Propulsion efforts were positively associated with shoulder positions but negatively associated with aerobic capacity, especially on the tile surface. However, muscle strength had the least impact on propulsion effort in five-minute propulsions. These results demonstrated that subjects with more rearward shoulder positions (in references to drive wheels) and better aerobic capacities tend to have a lower propulsion effort. However, the impact of aerobic capacity on propulsion effort became smaller on carpet compared to the tile surface. It is possible because that other factors, such as shoulder position, was more influential than physical fitness while maneuvering wheelchairs on a challenging surface.

Our finding in shoulder position was consistent with previous biomechanics research studying wheelchair axle position. According to Boninger‘s et al study [24], they found that the position of the axle relative to the shoulder at rest was related to biomechanical variables. Specifically, subjects with a further back shoulder position relative to the drive wheels exhibited decreased propulsion frequency and rate of rise of their propulsion force, but increased push angle [24]. Other studies also confirmed that a longer distance between the axle and backward shoulder position improved the push-time [23, 139], push angle [23, 139], and the smoothness of arm muscle activities [140]. From a biomechanical standpoint, an increased push angle would lead to more time to impart a force to the push-rim for reaching the same speed, and this would have the direct effect of decreasing propulsion frequency and force. These more favorable propulsion patterns would allow subjects to improve their economy by reducing propulsion efforts at the same speed.

Regarding the physical fitness of aerobic capacity and muscle strength, both factors did not significantly contribute to predicting propulsion efforts. However, aerobic capacity had a greater impact on propulsion effort than muscle strength. One possible reason is that our 5-minute wheelchair propulsion at moderate speeds is moderate in intensity. In other words, the energy to perform prolonged and moderate-intensity exercise comes primarily from aerobic metabolism, instead of anaerobic glycolysis and muscle power. According to Janssen's et al. [43] regression equations of physiological measurements, they found that only muscle isometric strength significantly contributes in predicting sprint power, maximal power output, and aerobic power. Zoeller et al. [41] also found the greater isokinetic elbow flexion and extension strength was associated with higher power output. Compared to previous studies, the exercise intensity in our study was ranging from 58±17 % (tile) to 70±21 % (carpet) peak VO_2. Therefore, the propulsion efforts measured through long and moderate-intensity propulsion were more indicative of aerobic capacity rather than active muscle mass.

Although physical factors did not significantly contribute to predicting propulsion efforts, the results do not dismiss the influence of physical fitness on daily activity. Instead, daily activity and exercise are known to reduce secondary conditions, especially for wheelchair users [141, 142]. According to Nooijen's et al. [141]longitudinal study, they found that peak oxygen uptake (peak VO_2) and peak power output were positively correlated with everyday activity level. In addition, stronger muscular strength could improve exercise performance by enabling higher levels of cardiorespiratory stress as the result of delayed local muscle fatigue [41]. Although physical activity is largely determined by factors that cannot be altered, such as lesion level, age, and gender, changeable factors such as activity level and body mass play an additional role [56]. Although our study results did not support the hypothesis that physical factors would influence propulsion efforts, it is important to understand that physical fitness does influence the general health and some types of exercise performance. Most important of all, the study highlighted that the biomechanical parameter, which is the relative distance between shoulder and drive wheels, is the most important operator factor that influences propulsion effort.

Aim 3.3 (mechanical parameters)

Several studies have looked at the effects of wheelchair designs, which included axle positions, tire types, and frame designs on propulsion performance. It is widely agreed that users propelling ultralightweight wheelchairs had a better performance and saved more energy than propelling standard wheelchairs [27, 98, 112]. Since there are many commercial designs and configurations of wheelchairs on the market, it is necessary to develop a protocol that can identify the impact of mechanical settings in a systematic way. To reach this goal, this study used the measurements of system mass, weight distribution, rolling resistance, and tire scrub as a systematic approach to quantify the mechanical properties of wheelchairs. In addition, these mechanical properties were fed into the regression model to evaluate the related influence of mechanical properties on wheelchair propulsion.

According to our regression models on both terrains, we found that only weight distribution significantly contributed to predicting propulsion effort. Subjects using wheelchairs with more loading on drive wheels required less propulsion effort. Although frictional parameters did not significantly contribute the model, by looking the sign of the coefficient, we are still able to see a trend that frictional parameters had a negative impact on propulsion efforts. In other words, subjects propelling wheelchairs with greater friction (more negative deceleration values) tend to need more propulsion effort.

There are two possible reasons that weight distribution influences propulsion effort so significantly. First, weight distribution has been shown to influence rolling resistance and tire scrub [28, 136]. Therefore, changing weight distribution will also coincide with friction changes that influence propulsion effort. Second, weight distribution is mainly affected by axle position, which has been widely studied to influence propulsion efforts [23, 24, 139, 140]. Therefore, the effect of weight distribution on propulsion efforts may be similar to the effect of axle position on propulsion efforts.

By looking into the correlation between weight distribution and frictional parameters, we found that weight distribution influences the overall friction in both straight and turning maneuvers but in opposing manners. However, the relationships were not clear in low friction condition (straight direction on tile) and high friction condition (turning direction on carpet). In addition to the weight distribution, we found drive wheel width would influence frictional parameters, especially during turns on the carpet. Overall, having a greater percentage loading on the wheelchair drive wheels increased deceleration while turning on a tile surface, but reduced deceleration while traveling straight on a carpet surface. The direction of the relationship between weight distribution and frictional parameters was consistent with the Aim 1.2 using an ISO dummy with different wheelchair configurations. In our over-ground maneuver, subjects were required to decelerate wheelchairs in the turning segment, but accelerate wheelchairs in the straight-line segment. Therefore, during freewheeling maneuvers, greater turning friction may be helpful for users to reduce their propulsion effort by facilitating deceleration, whereas lesser straight friction may be helpful for users to reduce their propulsion effort during acceleration.

Another interesting finding from this study is that system mass had no effect on propulsion efforts. The results were quite similar to previous findings that adding weight (less than 10 kg) to wheelchairs did not influence the propulsion kinetics [26, 143] and metabolic costs [27, 143] during the straight and steady maneuvers. Sagawa et al. [113] further found that additional mass (up to 5kg) had no influence on metabolic response during a series of ADL maneuvers. From a mechanical standpoint, wheelchairs with additional mass come with greater friction. Users therefore require a greater effort to overcome the friction for initiating the movement ($F=m\cdot a$). However, the human body might not be able to recognize the difference associated with small changes in wheelchair mass, especially during steady propulsion in a straight direction over a level surface. In addition, human body mass dominates the majority of wheelchair-system mass. According to our tested results, the wheelchair mass ranged from 14kg to 25kg, which only considered 11% to 26% of system mass. Referring to the Aim 1.2 results using ISO dummy, adding 5.5kg only increased rolling resistance by 2% and tire scrub by 11%. Therefore, the impact of mass on propulsion was not clear in this study. However, there is not doubt that wheelchair mass matters when users need to transfer wheelchairs manually in and out their cars.

The coding of Durable Medical Equipment (DME) by insurance carriers attempts to reflect the primary function of the equipment. In the case of manual wheelchairs, this is ease of propulsion. However, coding is not reflective of propulsion effort since no measures exist. Rather, the overall masses of wheelchairs are used to represent ease of propulsion. Wheelchairs are categorized using weight limits (i.e., 34 pound for lightweight wheelchairs (K0004) and 30 pounds for ultralightweight wheelchairs (K0005)). According to our study results, we found that weight distribution had much more impact than mass on propulsion effort. In other words, similar wheelchair mass but different weight distribution may cause significant differences in propulsion efforts. Therefore, the over-simplified approach of weight-classification could result in very disparate wheelchairs sharing the same category of coverage. This study’s findings will allow us to articulate differences and begin to reconcile these competing viewpoints.

Aim 3.4 (combined parameters)

This is the first study to explore the combined effect of mechanical and operator factors on propulsion efforts. The results of the present study indicate that the weight distribution of the wheelchair system had a greater influence on propulsion efforts than the physical fitness of the subject. Subjects using wheelchairs with more loading on drive wheels tended to reduce propulsion efforts, especially on the tile surface. Although it is not significant, subjects having a better aerobic capacity tended to have a reduced propulsion effort as well on tile surface. However, aerobic capacity had no impact on the carpet model since the standardized beta was near zero and R^2 actually decreased when it was added to the model. There are two possible reasons that could explain why weight distribution had a greater impact than physical fitness on propulsion effort. First, weight distribution was shown to correlate strongly with axle position and influence overall friction. Specifically, as more percent of the weight is placed over the larger rear wheels, this corresponds to an increasingly forward axle position and rolling resistance is decreased. These two factors were well proven to impact propulsion based on biomechanical benefits and reduced energy loss. Although the tire scrub would increase when more loading was placed on drive wheels, greater tire scrub could be beneficial for users to reduce their propulsion efforts by facilitating deceleration and turning. Second, our propulsion tasks, which simulate moderate-intensity daily maneuvers, could be less influenced by physical fitness than by propulsion skills and wheelchair configurations. Physical fitness may play a more important role when users are doing a high-intensity exercise, such as sprinting.

By comparing the regression model between the tile and carpet surface, we noted that the percentage of the propulsion effort variation that is explained by a linear model (R^2) dropped on the carpet surface. This finding highlighted that not only the weight distribution but other environmental factors influence propulsion efforts on the carpet surface. Although propelling wheelchair on the carpet was not hard for most participants (around 46% peak VO_2), propelling wheelchairs on the carpet (propulsion efforts: 9.9±2.3 ml/kg/min) was harder than tile surface (propulsion efforts: 7.3±1.8 ml/kg/min). By looking into the regression model of mechanical parameters on both surfaces, we found that the standardized beta of weight distribution drop on the carpet surface. Instead, the standardized beta of friction parameter raised. Friction parameter can be influenced either by surfaces, tire types (e.g. tire width, tire pressures, and tire material), or wheelchair designs (e.g. frame design and weight distribution). In addition, we found that shoulder position was more influential on the carpet than tile surface by looking into the regression model of operator parameters. Therefore, we can understand that not only the weight distribution but other factors, such as friction parameter and biomechanical benefits, become more important while increasing the difficulties of wheelchair maneuver.

4.2.4 Study limitations

The first limitation was that the small sample size for creating a multiple regression model, especially with two predictors. With a small sample size, the danger is a type II error. The limited sample size and injury types may induce a bias that compromises the homogeneous characteristic of subjects' functional capacity or wheelchair designs. Using a fixed model, R^2 deviation from zero, the power size tests were calculated from our main dependent outcomes, propulsion efforts for both SCI and AB groups. With the effect size (Cohen's f_2) ranging from 0.162 to 0.595, two predictors, and 36 samples, the power sizes were between 0.80 and 0.97. Specifically, the most powerful model was from mechanical predictors in the tile condition, whereas the least powerful regression model came from combined predictors (mechanical and operator factors) in the carpet condition. The possible reason for having a weak power size is that aerobic capacity (peak VO_2) was not normally distributed (Shapiro-Wilk p=. 001) but was positively skewed. In addition, 72% of our recruited subjects used a K0005 ultralightweight wheelchair. Therefore, the study would have a stronger result by having a greater sample size, especially if we recruited users with better physical fitness and wheelchairs of lower classification.

The second limitation is that the study included both able-bodied and full-time users in the data analysis. Able-bodied users may behave differently and use different propulsion skills compared to full-time users. Generally, people with SCI have lower energy expenditure than able-bodied individuals because of the reduced muscle mass and sympathetic nervous system available [144]. Users with different injury levels would also have different physical capacities [56] that influence exercise performance [51]. However, according to the objectives, this study was mainly designed to evaluate the impact of the operator and mechanical factors on propulsion efforts. As such, the inclusion of both populations is beneficial to have a generalized result of our research questions.

The third limitation is that our operator and mechanical parameters only explained 16% to 37% of the variance in propulsion efforts. In other words, our operator and mechanical parameters did not explain over half of the variance. One possible reason is that the familiarity with maneuvering wheelchairs, which includes propulsion skill and experience, might be the confounding factor that impacts propulsion efforts. Compared to the inexperienced users, the expert users generally present faster propulsion speeds, better energy efficiency technique, and more favorable propulsion patterns [98, 139]. In addition, inexperienced users might feel insecure while propelling wheelchairs first time in a foreign environment because of a lack of constant practice [98]. Consequently, increased intra-individual variability characteristics might bias our results. Another possible reason is that we did not include the impact of other biomechanical factors, such as backrest height, seat and backrest angle, the vertical position of the rear wheels, the camber of the rear wheels, cushion types, as well as frame and hand-rim design [98, 145]. In future studies, including propulsion experience, propulsion skills, and other biomechanical factors as predictors might help to build up the model by increasing the percentage of explained variance.

4.2.5 Conclusion

This study design provides a systematic approach to quantifying the mechanical properties of the wheelchair system. The multi-stage arm exercise and propulsion strength measurements also provide a valid and convenient approach to evaluating the aerobic capacity of wheelchair users. By designing a repeatable maneuver endowed with representative acceleration, stops, and turns, for the first time this study provides a direct approach to understanding the related influence of wheelchair designs and operator physical fitness on over-ground maneuvers. This study was able for the first time to find a related influence of operator and mechanical factors on propulsion efforts. According to our finding, the weight distribution of wheelchair systems was the most influential factor on propulsion efforts. Subjects propelling wheelchairs with more loading on drive wheels tend to need lesser propulsion efforts. However, the relationship was not clear when propelling on high friction surface, like the medium-pile carpet. Knowledge of how weight distribution impact wheelchair over-ground maneuvers are advantageous for clinicians and users when selecting or modifying a manual wheelchair.

CHAPTER 5
CONCLUSIONS

5.1 Summary findings

The overall goal of this project was to identify the impact of mechanical and operator parameters on wheelchair propulsion reflected in daily maneuvers. The central hypothesis was that both mechanical properties of wheelchairs and fitness level of operators would influence propulsion performance but on different levels. To complete this goal, test methods were developed or adapted to characterize the mechanical parameters of manual wheelchairs, measure biomechanical and physiological variables of wheelchair operators, and design an over ground propulsion task that reflects everyday mobility by requiring changes in speed and direction.

Our coast-down test methods can measure not only rolling resistance, but also tire scrub. This method of measuring resistive loss provides a simple and reliable approach. Through comparing different wheelchair configurations, we found that weight distribution has a greater impact on resistive losses compared to a 5.5 kg increase in wheelchair mass. In addition, the impact of weight distribution on resistive losses varies between rolling resistance and tire scrub. With more loading on the drive wheels, wheelchairs tend to have a less rolling resistance while traveling straight but greater tire scrub during turning.

Muscle strength related to the task of wheelchair propulsion is influenced by the upper extremity position and hand interface. The described method in measuring propulsion strength offers a reliable and valid method for assessing maximum isometric strength. This method can be easily applied in clinics to eliminate the need for wheelchair users to transfer out of their seats.

After all test methods had been developed and validated, wheelchair operators were recruited to propel over modified figure-8 course on both tile and carpet surfaces. This task was chosen because it involves changes in speed and direction, which better reflects the bouts of mobility performed in daily life. Propulsion effort was defined as the steady oxygen consumption during five-minute wheelchair propulsion using resting metabolism as the baseline correlation. A regression model was developed to identify the impact of the biomechanical and physical factors of operators as well as the mechanical properties of wheelchairs on propulsion efforts.

This is the first study to combine the assessment of inertial and frictional wheelchair parameters with operator variables in the study of propulsion effort. Weight distribution was the sole predictor of propulsion effort in both the mechanical parameter model and combined parameters (mechanical and operator) model. Shoulder position was the sole predictor in the operator parameters model. However, neither operator propulsion strength nor aerobic capacity significantly influenced propulsion effort during this sub-maximal propulsion task. The results show the significant influence of axle position on propulsion effort. Axle position impacts both weight distribution and the occupant's shoulder location relative to the drive wheels. Propulsion effort decreases as percentage weight is increased on the drive wheels and the shoulder becomes more aligned with the axle position. Weight distribution, however, becomes the most significant influence on propulsion effort when accounting for its relationship with shoulder position.

For wheelchair users, it is important to choose a wheelchair that can match their daily needs and anthropometric measurements. The customized wheelchair results in a wide variety of configurations, which includes seat angle, drive wheels, casters, and axle positions. Although this experimental design was not able to investigate specific configurations, the study was designed to evaluate the mechanical parameters of everyday wheelchairs systematically. The study results highlighted the importance that wheelchair axle position should be configured properly to improve propulsion performance by reducing physical stress.

5.2 Clinical relevance

The bulk of prior wheelchair propulsion studies assessed biomechanical and physiological variables using dynamometers, treadmills, and straight wheelchair maneuvers. These approaches allow for a more controlled assessment of biomechanics but are not able to study over ground propulsion that includes changes in speed and direction. The methodology used in this study was defined to begin an investigation into the complex relationships between mechanical wheelchair parameters and human variables [52].

A fuller understanding of the influence of wheelchair configuration on propulsion effort is vital for optimizing configurations for individual users. Because of the many commercial designs and configurations of wheelchairs, the results from previous biomechanical studies have limited the ability to compare wheelchair designs. By characterizing the inertial and frictional parameters of wheelchairs and then linking the related influence on propulsion effort, this study provides results that can serve as a useful reference to optimizing wheelchair configurations for individual users. For example, a wheelchair with a forward axle position (related short distance between the shoulder and axle positions) offers a biomechanical advantage and a decrease in friction during straight trajectories. However, it is unclear whether wheelchair users will be afforded the same benefits during daily maneuvers, which include turning, since frictional loss from tire scrub is elevated. Another feature of this study is that the results, which include the effects of momentum changes, reflect how MWUs use their wheelchairs in daily life. The modified figure-8 course more effectively accounts for users' daily activities than previous treadmill/roller studies. In other words, this study provides an overall assessment of wheelchair design, especially during over-ground maneuvers. It also evaluates the impact of individual fitness on propulsion efforts.

In general, this study does not seek to draw inferential conclusions about dichotomous comparisons between wheelchair configurations. Instead, we used the everyday wheelchairs of wheelchair users to consider the variety of wheelchair configurations, which reflected in different wheelchair frames, tires, and seats. This study provided a systematic approach to quantify the mechanical properties of wheelchairs and the physical fitness of operators. This study further employed regression modeling to determine the related influences on propulsion efforts resulting from inertia and friction of the wheelchair as well as the biomechanics and physical fitness of operators. This study design distinctively seeks to enhance the applicability of the findings to the clinical practice of configuring wheelchairs for specific users and for the particular manner in which they use wheelchairs. Because propulsion effort is related to inertial and frictional properties rather than limited to specific wheelchair models, we can accomplish this goal and thus provide more thorough knowledge about the manner in which these parameters impact propulsion efforts.

5.3 Future direction

There are a number of questions that need further study based on the study finding. First, it will be clinically important to study which configuration has the most impact on user performance. Second, it is vital to consider both stability and maneuverability while changing wheelchair weight distribution. Third, future wheelchair studies should include different populations to benefit more user groups. Last, utilizing activity monitors can be an alternative approach to quantifying the performance of daily propulsion, which includes different terrains and propulsion patterns.

To look back at the first question, we understand that wheelchair optimization is a big challenge. Wheelchairs can be configured with different parts based on user needs and the maneuverability requirements of various environments. Even within the same wheelchair, several modifications are needed to match user anthropometrics [109]. This study design provides an alternative approach to describing the impact of wheelchair designs and human bodies on wheelchair propulsion in a systematic way. However, the study was not designed to explain the direct impact of each configuration on propulsion performance. The current wheelchair market offers a huge variety of components, all of which can be combined to create a seemingly endless number of wheelchair configurations. Designing a human subject-based experimental protocol to test the impact of all these configurations is not feasible for multiple reasons. For example, if we wanted to evaluate the interaction effects of two axle positions, two frame designs, and two tire types on propulsion performance, participants would need eight times the number of repeated measurements to test each configuration. In other words, studying more wheelchair factors will extend the protocol significantly, resulting in increased psychological and/or physical fatigue for the participant. In addition, changing wheelchair configurations would also be problematic. If using user wheelchairs, researchers would face the dilemma that the majority of wheelchairs are not adjustable. If experimental wheelchairs are used to circumvent this challenge, participants would still need time to acclimate to the test wheelchair. To overcome these challenges, developing a robot-operated system [146] will be helpful since it has the potential to make many more comparisons than human subject testing could achieve. Users bring more variability (and reduced internal validity) due to differences in posture and propulsion biomechanics, but also bring more applicability (and enhanced external validity). Therefore, using a robot-operated system can be a good start to compare several configurations with precise and reliable measurements. Human-based studies can be further used as secondary comparisons to confirm whether the difference is clinically relevant.

Second, wheelchair static and dynamic stability is another topic that needs further study. By definition, wheelchairs are stable as long as the system’s center of gravity stays inside the wheelchair’s base of support, which is generally defined by where the wheels contact the ground. According to the study results, we have shown that weight distribution would influence overall friction and also propulsion effort. More loading on drive wheels (forward axle position), led to less rolling resistance and thus reduced user propulsion efforts. However, manual wheelchairs with the forward axle position would also reduce the rear stability and cause backward tipping [147]. Previous epidemiology studies have indicated that the majority of serious wheelchair user injuries came from tipping or falling out of wheelchairs [148-150]. According to Gaal's et al. study [147], directions of falls were associated with wheelchair types and riding surfaces. For example, wheelchairs with less percent loading on drive wheels could result in decreased maneuverability and increased downhill turning tendency. By definition, the downhill turning tendency is approximately proportional to the normal horizontal distance from the center of mass to the drive wheel axis [61]. Wheelchairs would also tip forward easily when wheelchairs were slowed down or stopped by the terrain, such as a low curb. In contrast, wheelchairs with more percent loading on drive wheels could tip backward easily, especially while riding uphill. Therefore, finding an ideal wheelchair weight distribution that considers the balance between maneuverability and stability would be necessary for clinical wheelchair prescriptions. In addition, the ergonomics of propulsion must also be considered while changing weight distribution by moving drive wheel axle position.

Third, including different wheelchair populations in the study would benefit the generalizability of the results. Within our study participants, all full-time wheelchair users were diagnosed with SCI. According to Hubbard's et al. study [151] in analyzing the national database, 45% of SCI and 18% traumatic brain injury (TBI) patients using manual wheelchairs used either lightweight (K0004) or ultralightweight (K0005) wheelchairs, but SCI and TBI are not the major diagnoses among those who use wheelchairs [151]. Instead, COPD/CHF (chronic obstructive pulmonary disorder/chronic heart failure) (23%) was the most frequent primary diagnosis of veterans who received the wheeled mobility equipment, followed by stroke (16%) and arthritis (11%). In addition, patients with different diagnoses would have different considerations while selecting and configuring wheelchairs. For example, stroke patients may use a foot propulsion technique that uses both their unaffected arm and leg to propel their wheelchairs. This type of propulsion technique requires different muscle groups and patterns compared to traditional two-hand propulsion [152]. Compared to two-hand propulsion, users with foot propulsion tend to deviate to the hemiparetic sides, thus resulting in more propulsion strokes and slower propulsions speeds [152]. As a result, future studies need to include various cohorts instead of targeting SCI users to improve the process of wheelchair prescriptions and training.

Finally, future studies need to consider the real situations that users encounter in daily life. Physical activity monitors can be an effective indication of wheelchair usage and exercise intensity involved in the daily activities. According to our research questions, we measured metabolic efforts from five-minute propulsions around a modified figure-8 course on tile and carpet surfaces. Although we considered the variance in propulsion patterns (e.g. change of speed and direction) and floor friction (e.g. tile and carpet), our study has limitations in explaining certain terrain designs (e.g. slope and cross-slope) and short-bout activity (e.g. bouts 8 meters in propulsion distance and lasting less than 20 seconds [153]). Therefore, there is a need to consider the real propulsion condition that users encounter in daily lives. Previous wheelchair studies have investigated the impact of indoor and outdoor terrains on propulsion performance, such as over cross-slope [61, 147, 154], ramps [22, 58, 155], and grass [58]. It has been shown that users propelling a typical wheelchair on a 2-degree cross-slope spent about twice the effort as propelling on a level surface [61] due to increased push strokes and higher push force [154]. Users also needed one and two times greater start-up forces when propelling on a 5^o ramp and grass, respectively, versus a level tile surface [58]. According to previous studies, the differences between surfaces in required propulsion efforts highlight the importance of evaluating wheelchair propulsion ability over a range of surfaces. However, it is a challenge to mimic a real-world environment that MWUs encounter every day. Therefore, using a portable monitor became a common approach to track users' physical activities on a daily basis [156]. Previous studies have applied accelerometer-based activity monitors on upper extremities to track daily life activities [157, 158]. With respect to wheelchair mobility, previous researchers have placed a wheel rotation data logger [159], gyroscope [158], and tri-axial accelerometer [153, 160] on wheelchair wheels to track wheel revolutions, direction, and duration of movement in community settings. Thanks to the progress of technology, wearable sensors are more reliable and accurate than ever before. Therefore, the availability of activity monitors can potentially help clinicians and researchers to understand the users' physical activity and energy expenditure on a daily basis. The technique in measuring daily activity can be further applied to understanding whether the improvement of wheelchair designs or operator fitness levels would lead to a healthier and more active lifestyle.

Wheelchair users are quite diverse but they share a functional similarity: they use wheelchairs to assist in mobility. For many manual wheelchair users, wheelchairs are used both within the home and community. As a result, community living and participation are directly impacted by the ability of a wheelchair to facilitate independent mobility. Propelling wheelchairs with decreased effort may permit additional functional capacity as well as reduced loading on the upper extremities. By extension, the selection of a wheelchair that properly meets the functional goals of the user is important for facilitating independence, participation, and quality of life. However, the selection of wheelchair type, components, and configurations involves negotiating a series of compromises, because these choices have tradeoffs in performance, function, cost, complexity, and a host of other factors. With more scientific knowledge of wheelchair studies about propulsion effort, users would become more empowered to make choices reflective of their needs and desires. Manufacturers would also benefit by having a valid means to inform the design of new wheelchairs.

APPENDIX A

THE MULTISTAGE SUBMAXIMAL EXERCISE PROTOCOL

Preparation

1. The participant should not have a meal, or any caffeine drinks less than 3 hours before the exercise.
2. Medications are documented.

Procedure

1. With the participant seated in a wheelchair, obtain resting heart rate
2. Calculate maximum HR = (220-age)
3. Calculate 90% submaximal HR= (0.90* ((220-age) -HRrest)) +HRrest. This value will be used as a cutoff reference.
4. Set participant up with the arm crank device (the pedal axis is at shoulder height). Make sure that the participant's wheelchair is positioned at the table so that slightly less than full elbow extension would occur at the furthermost point in the pedal's revolution range.
5. Start the test by having the participant cycle at a 70rpm (±5 rpm) for 3 minutes (10W).
6. The participant will have at least 1-minute rest after each stage. During the resting period, subjects will continually paddle the arm crank with low speed (35 rpm) and low resistance (5W)
7. Record the RPE (6-20 scales) right after 3 minutes of exercise. An operator will ask "how hard you feel about the intensity of this arm exercise?"

8. Add a 10-Watt resistance for each stage of exercise (maximum resistance: 30W) and maintain cycling at 70 rpm for 3 minutes.
9. Continue cycling and monitor metabolic response until the subject desires to stop or after three stages of resistance, whichever comes first.
10. If subjects have recovered after 5 minutes and do not show or report any signs of distress, the test may be terminated.
11. If the participant shows or reports signs of distress, researchers will wait until the participant has fully recovered from the test before discharging the participant from the session.

The endpoint will be defined when the subject reaches the three-stage exercise or has one of the following responses:

1. Cranking rate is below 60rpm
2. RPE >=19
3. >= 90% reserved HR max
4. Chest pain
5. Dyspnoea
6. Volitional fatigue
7. Angina

APPENDIX B

CHARACTERIZATION OF PEOPLE WITH AND WITHOUT SCI

Table B.1 The description of metabolic measurements for able-bodied group

Subject #	**Age**	**Sex**	**Resting VO_2** [ml/kg/min]	**Aerobic capacity** [ml/kg/min]	**Net effort_Tile** [ml/kg/min]	**Net effort_Carpet** [ml/kg/min]
1	19	F	4.1	29.8	3.6	8.1
2	20	M	5.4	23.8	6.0	11.8
3	21	M	4.4	25.0	6.5	9.4
13	26	M	4.5	18.1	6.3	9.4
19	22	F	4.1	27.8	10.2	10.7
20	47	M	3.4	14.4	7.6	13.6
22	19	F	3.6	20.6	7.4	10.0
23	20	M	4.0	20.6	9.1	14.0
24	24	M	3.2	15.8	9.4	10.6
25	21	F	3.9	19.7	8.9	11.9
26	24	M	3.5	20.5	5.0	6.9
27	23	F	5.0	18.5	8.6	10.8
28	24	M	5.3	21.2	7.8	10.1
Mean ± Standard Deviation				F (n=5): 23.3 ± 4.6 M (n=8): 19.9 3.4		

F: female; M: male

Our results showed that able-bodied subjects (4.18± .71 ml/kg/min) had a significantly higher resting metabolic rate compared to SCI subjects (3.49± 1.01 ml/kg/min), $t\ (28)= 2.089$ p=0.046. It is possible that the able-bodied group (24±7 years old) was much younger than the SCI group (40±10 years old). In addition, wheelchair users with SCI had a lower fat-free mass due to the injury.

Our predicted peak VO_2 from the able-bodied group had a lower value for the male group but higher value for a female group compared to Al-Rahammneh's et al. [127] study using arm-crank ramp exercise. According to Al-Rahammneh's et al. study, nine able-bodied men had an averaged peak VO_2 with 28.4 ± 5.0 ml/kg/min, whereas seven able-bodied women had an averaged peak VO_2 with 19.0 ± 2.2 ml/kg/min. The peak VO_2 for the able-bodied group also had a similar value compared to Hooker's et. al [161] study using arm crank ergometer (N=15, peak VO_2 = 24.5± 4.5).

Table B.2 The description of metabolic measurements for SCI group

Subject #	**Age**	**Sex**	**Injury level**	**Resting VO_2** [ml/kg/min]	**Aerobic capacity** [ml/kg/min]	**Net effort_Tile** [ml/kg/min]	**Net effort_Carpet** [ml/kg/min]
4	28	M	T1	3.3	16.3	9.0	10.7
5	46	M	T1	3.2	10.3	4.6	6.1
6	46	M	T4	2.8	16.5	4.9	8.3
8	24	M	T5	3.6	21.3	9.2	11.1
9	29	F	C1	2.4	16.9	7.3	9.9
10	46	M	T4	3.1	17.5	10.3	13.2
12	46	M	T4	2.1	13.7	8.2	11.1
14	47	M	C7	3.4	12.1	4.9	4.7
15	34	M	T12	3.3	18.0	8.7	11.9
16	32	M	T3	3.3	18.2	6.6	12.5
17	58	M	T12	3.7	36.0	5.7	6.5
18	48	M	T9	2.5	17.4	6.3	8.0
21	46	M	T9	2.9	22.8	6.9	10.4
29	57	M	C5	4.8	13.2	6.1	6.9
30	36	M	T12	4.9	48.9	4.0	6.9
31	54	F	L1	2.7	22.3	7.1	8.1
32	35	M	C5	3.5	12.1	6.0	7.5
33	43	M	T5	2.6	21.7	10.3	9.6
34	31	M	C5	5.0	21.2	5.5	5.9
35	33	M	T12	5.6	30.9	7.3	8.8
36	28	M	T5	3.4	18.7	9.0	13.9
Mean ± Standard Deviation					Low lesion (T7-T12, n=14): 26.9 ± 11.1 ml/kg/min High lesion (T6 above, n=7): 16.2 ± 3.8 ml/kg/min		

F: female; M: male

Our predicted peak VO_2 from SCI group had a wider range (from 12 to 49 ml/kg/min) compared to Yamasaki's et al. [162] and Al-Rahammneh's et al. [163] studies using arm-crank ramp exercise. According to Yamasaki's et al. [162] study of evaluating metabolic responses in people with paralysis, 28 males (16 subjects were active athletes) with paraplegia (injury level: T3 to L3; age: 21 to 52 years old) had a peak VO_2 ranging from 17.6 to 30.9 ml/kg/min. Al-Rahammneh's et al. [163] also had a similar result that 5 active SCI men subjects (injury level: T6 to L1; age: 22 to 31 years old) showed a peak VO_2 ranging from 22 to 34 ml/kg/min. By separating the SCI lesion level, we had a similar peak VO_2 in high lesion group but a higher peak VO_2 in low lesion group compared to Hooker's et. al [161] study (high lesion (N=13): 17.9±4.1 ml/kg/min; low lesion (N=14): 20.6±4.6 ml/kg/min).

By looking into the value of propulsion efforts (resting + net propulsion efforts) from both able-bodied and SCI group, we noted that some of the participants had a greater oxygen consumption while propelling on carpet than predicted aerobic capacity. It is possible that the predicted peak VO_2 by RPE approach may underestimate the participants' real aerobic capacity. For example, participants may report high RPE because of localized fatigue (muscle sore) instead of cardiorespiratory stress while doing arm activity. The localized fatigue can be influenced by injury level, the experience in arm activities, or the strength of upper extremities.

APPENDIX C

PREDICTED PEAK VO_2 FROM THE RESPONSE OF HEART RATE DURING SUBMAXIMAL ARM EXERCISE

Maximum oxygen consumption can be estimated from heart rate (HR) responses to submaximal exercise by extrapolating the relationship to the subject's age-adjusted estimate of maximal heart rate (220-age). It is one of the common procedure used with a graded exercise test in a healthy population. This technique provides the features of low cost, ease of measurement, and high reliability. The purpose of this **Appendix C** is to compare the difference of peak VO_2 prediction using RPE and HR.

Table C.1 The aerobic capacity predicted by RPE and HR for the able-bodied group

Subject #	Age	Sex	Aerobic capacity (RPE) [ml/kg/min]	Aerobic capacity (HR) [ml/kg/min]
1	19	F	29.8	24.9
2	20	M	23.8	27.9
3	21	M	25.0	38.7
13	26	M	18.1	36.0
19	22	F	27.8	28.3
20	47	M	14.4	24.6
22	19	F	20.6	24.5
23	20	M	20.6	24.9
24	24	M	15.8	23.7
25	21	F	19.7	23.1
26	24	M	20.5	23.6
27	23	F	18.5	22.0
28	24	M	21.2	35.4

F: female; M: male

Table C.2 The aerobic capacity predicted by RPE and HR for SCI group

Subject #	Age	Sex	Injury level	Aerobic capacity (RPE) [ml/kg/min]	Aerobic capacity (HR) [ml/kg/min]
4	28	M	T1	16.3	16.74
6	46	M	T4	16.5	33.16
8	24	M	T5	21.3	19.19
9	29	F	C1	16.9	15.42

10	46	M	T4	17.5	15.80
12	46	M	T4	13.7	13.21
14	47	M	C4	12.1	16.7
15	34	M	T12	18.0	16.5
16	32	M	T3	18.2	34.2
17	58	M	T12	36.0	37.7
18	48	M	T9	17.4	17.1
21	46	M	T9	22.8	24.1
29	57	M	C5	13.2	26.4
30	36	M	T12	48.9	37.4
32	35	M	C5	12.1	25.1
33	43	M	T5	21.7	18.8
34	31	M	C5	21.2	69.9
35	33	M	T12	30.9	40.6
36	28	M	T5	18.7	18.2

F: female; M: male

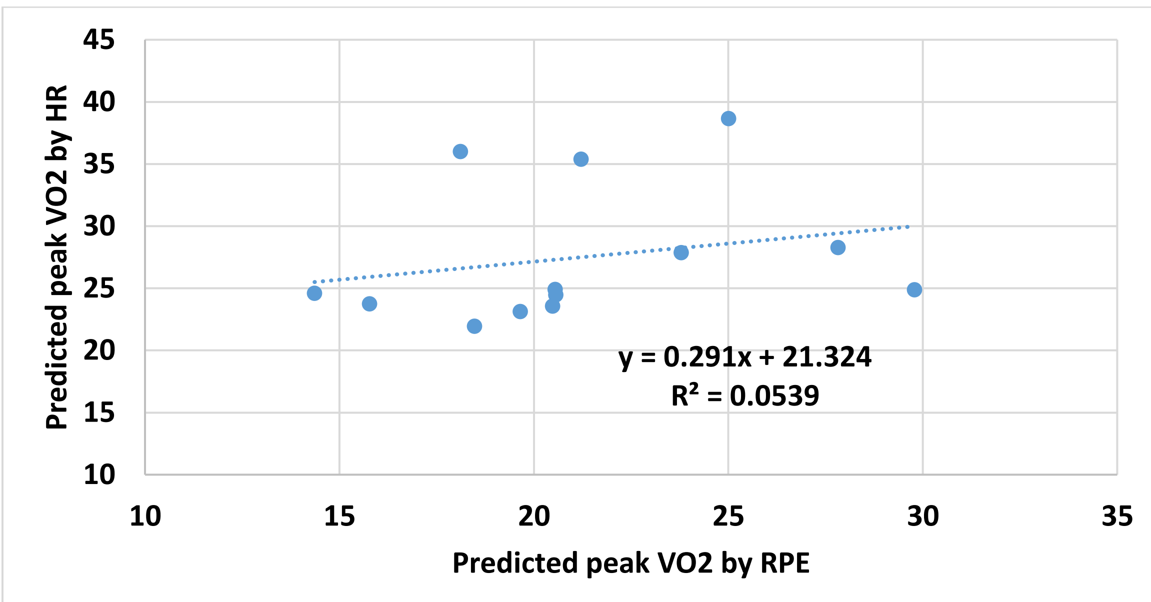


Figure C.1 The correlation between RPE and HR methods in predicting peak VO_2 for able-bodied group

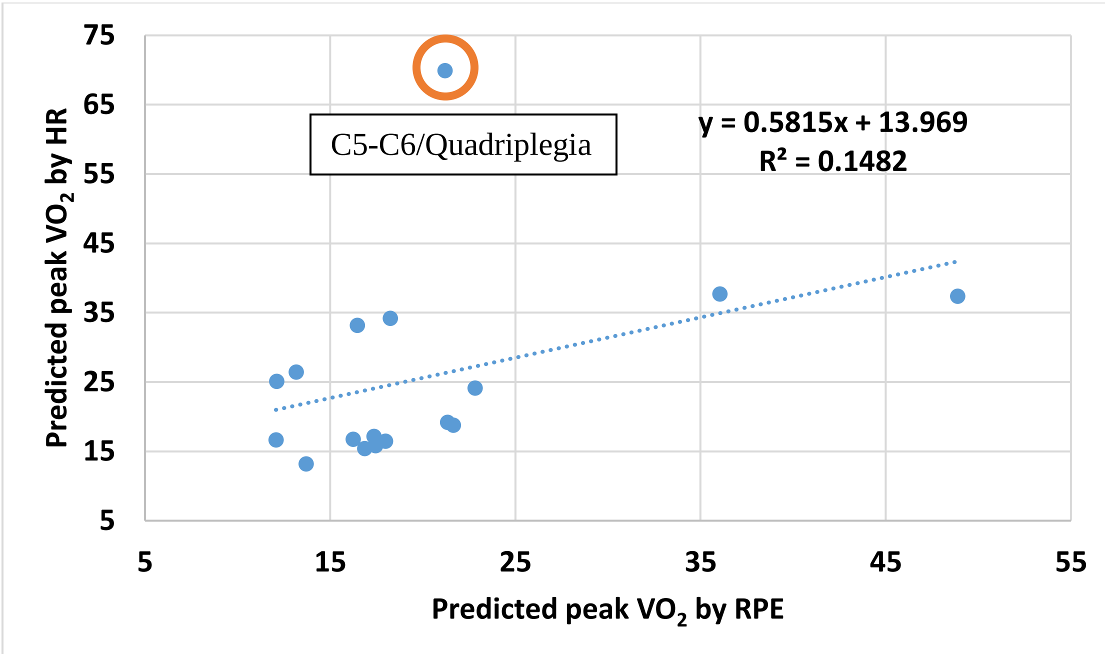


Figure C.2 The correlation between RPE and HR methods in predicting peak VO_2 for SCI group

Our results did not show a high correlation between RPE and HR prediction in both able-bodied and SCI populations. Consistently, the HR method (able-bodied: 29.1±11.9; SCI: 25±10 ml/kg/min) predicted a greater peak VO_2 than the RPE method (able-bodied: 19.9± 4.8; SCI: 23±10 ml/kg/min) in both populations. However, only able-bodied group demonstrate a significant difference between two methods, t(16)= -3.154, *p*= .01. The first possible reason is that several variables that affect submaximal heart rate will affect the slope of the HR/VO_2 line and thus the estimated peak VO_2. These variables include eating before the test, hydration, body temperature, emotional state of the subject, medications. For subjects with SCI, HR response can even be influenced by injury type and injury level. According to Lewis's et al. [164]study, they found that subjects with tetraplegia had a lesser increase in heart rate than subjects with quadriplegia while increasing work load, which is reflected in increased RPE.

The second possible reason is that arm cranking exercise solely relies on the upper extremity muscle. In other words, subjects could limit their exercise performance because of fatigue or muscle sore instead of cardiovascular stress [164]. The able-bodied population might have more influence of arm exercise than SCI population since able-bodied population usually do not do arm activity as intense as SCI population. Third, the estimation of maximum HR can be influenced by other variables rather than age. Londeree et al. [165] found that age accounted for about 70-75% of the variability in predicting maximum HR. Other factors, such as gender, the level of fitness, type of ergometer, exercise protocol, and the race could consider another 20-25% of the variability in predicting maximum HR.

APPENDIX D

OTHER POSSIBLE OPERATOR PREDICTORS

Table D.1 Data description (n=36)

Propulsion task			**Multistage arm ergometry exercise**			
	Total oxygen cost [ml/kg/min]	**RPE**		**10 W**	**20 W**	**30 W**
Tile	10.7±1.7	11±2	**Oxygen consumption** [ml/kg/min]	7.1±1.4	11.0±2.6	13.8±4.5
Carpet	13.4±2.6	13±2	**RPE**	10±2	13±1	15±2

Mean±SD; RPE: Rated Perceived Exertion

Table D.2 Spearman correlation

			Metabolic cost [ml/kg/min]		
	Aerobic capacity (RPE)	Aerobic capacity (HR)	10 W	20W	30W
Net effort_Tile	-.176 (p=.311)	**-.490* (p=.004)**	**-.482* (p=.003)**	-.266 (p=.122)	-.295 (p=.085)
Net effort_Carpet	-.054 (p=.760)	**-.453* (p=.008)**	-.162 (p=.352)	.025 (p=.885)	.126 (p=.472)
Aerobic capacity (RPE)		**.457* (p=.007)**	**.537* (p=.001)**	**.711* (p=.000)**	**.633* (p=.000)**
Aerobic capacity (HR)			**.560* (p=.001)**	**.521* (p=.002)**	**.415* (p=.015)**

**p*<.05, Aerobic capacity (RPE) and (HR) is the peak VO_2 predicted by RPE and heart rate, respectively.

Table D.3 Partial correlation (control variable: shoulder position, XPOS)

			Metabolic cost [ml/kg/min]		
	Aerobic capacity (RPE)	Aerobic capacity (HR)	10 W	20W	30W
Net effort_Tile	**-.362* (p=.046)**	-.325* (p=.074)	**-.462* (p=.009)**	-.277 (p=.131)	**-.404* (p=.024)**
Net effort_Carpet	-.191 (p=.302)	**-.367* (p=.043)**	-.030 (p=.872)	.008 (p=.967)	.040 (p=.831)
Aerobic capacity (RPE)		.254 (p=.168)	**.537* (p=.002)**	**.637* (p=.000)**	**.386* (p=.032)**
Aerobic capacity (HR)			**.362* (p=.045)**	**.356* (p=.049)**	.211 (p=.255)

**p*<.05

Table D.4 Spearman correlation

	RPE		
	10 W	20W	30W
Net effort_Tile	-.098 (p=.574)	.003 (p=.985)	.113 (p=.525)
Net effort_Carpet	-.173 (p=.321)	-.008 (p=.964)	.126 (p=.478)
Aerobic capacity (RPE)	-.073 (p=.672)	-.203 (p=.234)	-.288 (p=.093)
Aerobic capacity (HR)	.251 (p=.152)	.267 (p=.127)	.100 (p=.579)

The purpose of **Appendix D** is to investigate whether other metabolic factors (RPE and oxygen consumption) measured by arm ergometry can improve the prediction of propulsion efforts. By comparing the metabolic efforts (oxygen consumption, VO_2 and self-reported exertion, RPE) (**Table D.1**) between propulsion tasks and arm ergometry exercise, participants propelling wheelchairs on tile had a similar VO_2 to arm exercise with 20W resistance, whereas propelling wheelchairs on the carpet had a similar VO_2 to arm exercise with 30W resistance. However, RPE in tile condition was similar to arm exercises with 10W, whereas RPE in carpet condition was similar to arm exercises with 10W.

By looking the nonparametric correlation between dependent variables (net propulsion efforts on tile and carpet) and metabolic measurements from arm ergometry exercise, we noted that net propulsion effort on tile was significantly correlated with VO_2 with either 10W or 30W arm resistance (**Table D.2**). However, by controlling the effect of shoulder position (shoulder position was the only operator variable that added statistically significantly to the prediction model), only VO_2 with 10W arm resistance was significantly correlated with net propulsion effort on the tile (**Table D.3**). Compared to tile condition, net propulsion efforts on the carpet was not correlated with VO_2 either with 10W, 20W, or 30W arm resistance. In addition, there was no correlation between net propulsion efforts and RPE measured during arm exercise (**Table D.4**).

Other than predicted aerobic capacity, metabolic cost with 10W arm resistance could be another physiological factor to predict propulsion effort. Arm exercise with 10W resistance had lower oxygen consumption but similar RPE compared to wheelchair propulsion on the tile. However, by comparing the correlation coefficient to mechanical (weight distribution) and biomechanical parameters (shoulder position), physiological parameters still had the lowest value. Therefore, different physiological factors did not change the related influence of mechanical and biomechanics parameters on propulsion efforts.

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VITA

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